**Universidade de São Paulo
Faculdade de Medicina de Ribeirão**

2021

**Aplicação de um modelo musculoesquelético para análise
biomecânica da articulação do joelho: efeito agudo da fadiga em
atletas de handebol feminino**

PPGRDF
PROGRAMA DE PÓS-GRADUAÇÃO EM
REABILITAÇÃO E DESEMPENHO FUNCIONAL
FMRP-USP

Bruno Luiz de Souza Bedo

Tese

Versão corrigida. A versão original encontra-se disponível tanto na Biblioteca da Unidade que aloja o Programa, quanto na Biblioteca Digital de Teses e Dissertações da USP (BDTD)

BRUNO LUIZ DE SOUZA BEDO

**Aplicação de um modelo musculoesquelético para análise
biomecânica da articulação do joelho: efeito agudo da fadiga em
atletas de handebol feminino**

Documento apresentado a Faculdade de Medicina de Ribeirão Preto da Universidade de São Paulo como parte dos requisitos exigidos para obtenção do título de Doutor em Ciências, na área de Reabilitação e Desempenho Funcional.

Área de concentração: Educação Física

Orientador: Prof. Dr. Paulo Roberto Pereira Santiago

Coorientador: Prof. Dr. Renato de Moraes

Ribeirão Preto
2021

Autorizo a divulgação parcial ou total destes estudos, por qualquer meio convencional ou eletrônico, para fins de estudo e pesquisa, desde que citada à fonte.

Faculdade de Medicina de Ribeirão Preto da Universidade de São Paulo

Bedo, Bruno Luiz de Souza.

Aplicação de um modelo musculoesquelético para análise biomecânica da articulação do joelho: efeito agudo da fadiga em atletas de handebol feminino. / Bruno Luiz de Souza Bedo; orientador Paulo Roberto Pereira Santiago; coorientador Renato de Moraes – São Paulo, 2021.

196 f.: il.

Tese (Doutorado) – Universidade de São Paulo, 2021.

Biomecânica do joelho. 2. Fadiga. 3. Simulação e Modelagem Computacional. I. Santiago, Paulo Roberto Pereira. II. Moraes, Renato de. III Bedo, Bruno Luiz de Souza. Título: Aplicação de um modelo musculoesquelético para análise biomecânica da articulação do joelho: efeito agudo da fadiga em atletas de handebol feminino.

BEDO, Bruno Luiz de Souza

Aplicação de um modelo musculoesquelético para análise biomecânica da articulação do joelho: efeito agudo da fadiga em atletas de handebol feminino.

Documento apresentado a Faculdade de Medicina de Ribeirão Preto da Universidade de São Paulo como parte dos requisitos exigidos para obtenção do título de Doutor em Ciências, na área de Reabilitação e Desempenho Funcional.

Aprovado em:

Banca examinadora

Prof. Dr. **Felipe Pivetta Carpes**

Julgamento: **Aprovado**

Instituição: **UNIPAMPA– Campus Uruguaiana**

Assinatura_____

Profa. Dra. **Anamaria Siriani de Oliveira**

Julgamento: **Aprovado**

Instituição: **FMRP-USP – Ribeirão Preto**

Assinatura_____

Prof. Dr. **Luis Mochizuki**

Julgamento: **Aprovado**

Instituição: **EACH-USP – São Paulo**

Assinatura_____

DEDICATÓRIA

Aos progenitores, Mario Luiz Bedo Junior e Maria Ruth de Souza Bedo, que me oferecem amor incondicional e incansáveis oportunidades de aprendizado. Vocês são os maiores exemplos de companheirismo, compromisso e dedicação. Hoje, o sonho que sonhamos juntos, concretiza-se.

Ao meu melhor amigo e maior exemplo, o irmão Douglas Bedo, por me ensinar a enxergar a alegria na simplicidade das coisas. A sua força e perseverança me fortalecem o desejo de ser alguém melhor.

Aos avós paternos, Mário (*in memoriam*) e Neide, por me ensinarem a compreender que a verdadeira felicidade não se encontra em coisas palpáveis, e que a família é o bem mais precioso de um homem. Obrigado por me cercarem de carinho, amor e experiências que me trilharam para o caminho do bem.

Aos avós maternos (*in memoriam*) Deoreste e Ruth que, embora por um curto período de tempo, me ofereceram o mais puro amor e cuidado.

AGRADECIMENTOS

A vida, o maior dom a nós foi concebido, intriga gerações e até hoje não foi integralmente compreendida. Questiona-se a maneira que teve inícios e, claramente, o que acontece após o seu término físico. No entanto, mais importante do que compreender o seu início e fim, faz-se necessário refletir o que dela nós fazemos. Deste modo, o livre arbítrio possibilita-nos colocar em prática qualquer verbo que desejemos. Mas embora seja uma das dádivas mais belas, o ato de agradecer frequentemente é banalizado. Portanto, para vocês, amigos e familiares, abaixo descritos, deixo os meus mais sinceros e profundos agradecimentos. Sem vocês, meus caros, absolutamente nada disso seria possível.

O maior egoísmo de um cientista é asseverar que a ciência é a única verdade. Há situações que a ciência não basta e, nesses momentos, nos sustentamos na divindade. Assim, agradeço ao grande arquiteto do universo, **Deus**, pelo dom da vida. Pela oportunidade de ser alguém melhor diariamente e, principalmente, por afugentar os medos nos momentos de escuridão.

Perante os privilégios que a mim foram concebidos ao decorrer da vida, destaco a base familiar que tive e tenho comigo. Agradeço aos meus primeiros mentores, meus exemplos, **Mario Luiz Bedo Junior** e **Maria Ruth de Souza Bedo**. Dentre os diversos ensinamentos, ressalto as prioridades para um homem: família e educação. A primeira por serem as pessoas em quem você deve/pode cegamente confiar. A segunda por ser algo que absolutamente ninguém pode lhe tomar. Assim, Pai e Mãe, obrigado pela insaciável disposição de oferecer a melhor educação para os filhos. Mesmo que, em algumas situações, faltasse-nos o superficial, a educação jamais foi posposta. Prometo a vocês a gratidão eterna, amor e cuidado incondicional. A você, meu melhor amigo, **Douglas Bedo**, obrigado pela paciência, exemplos e pureza. Você, meu caro, é um guerreiro do qual eu me orgulho imensamente. Amo vocês.

Agradeço ao orientador que, com o passar dos anos, se tornou um grande amigo, **Dr. Paulo Roberto Pereira Santiago**. A você, *Preto*, agradeço a oportunidade de aprender contigo. Obrigado por enxergar um potencial que, muitas vezes, particularmente, nem eu mesmo encontrava. Agradeço, principalmente, por não medir esforços pessoais e profissionais para oferecer todo conhecimento. Obrigado por abrir portas que eu jamais seria capaz de abrir sozinho. Você, meu amigo, é um grande exemplo.

Durante a minha caminhada na pós-graduação, frequentemente (re)afirmei que nunca tive um co-orientador mas, sim, dois orientadores. Deste modo, **Dr. Renato de Moraes**, agradeço a possibilidade de aprender com o senhor. A sua personalidade serena, íntegra e transparente, nos fazem acreditar que em meio à uma sociedade egoísta, ainda existem profissionais éticos. Agradeço por me acolher como aluno. Por me oferecer uma bolsa de pesquisa mesmo sem eu ser diretamente um aluno seu que, independentemente do valor financeiro, acrescentou muito conhecimento e experiência profissional. Agradeço pelas reuniões praticamente semanais que, com o passar do tempo, se tornaram bate papos para conselhos profissionais. Obrigado pela confiança e a oportunidade de trabalhar contigo.

Agradeço fortemente ao companheiro de trabalho e amigo **Dr. Danilo Catelli**. Obrigado pelas possibilidades que você me ofereceu. Por me apresentar mares que eu jamais desbravaria sozinho. Agradeço o suporte que você e sua bela família me ofereceram durante a minha estadia no Canadá. Você, meu amigo, é um dos exemplos pelo qual o nosso país deve se orgulhar de seus pesquisadores. Obrigado e espero que logo nossos caminhos profissionais se cruzem novamente. *Thank you.*

*I would like to thank my Canadian supervisor Prof. Dr. **Mario Lamontagne** at the University of Ottawa, for accepting my Ph.D. internship. I also thank you for giving me unconditional support during my time in Canada. Thank you for trusting in my capacity and for all knowledge*

shared. Lastly, I appreciated each research and particular story during lunchtime! Merci, Mario.

Agradeço aos meus (ex) companheiros do Laboratório de Biomecânica e Controle Motor: **Rodrigo Aquino, Vitor Marquetti, Luiz Henrique Palluci, Rafael Akira, Nilson Ribeiro, Marina Villalba, Matheus Pacheco, Luciana Oliveira, Andressa Busch, Andreia Abud, Rosângela Batistela e Eduardo Bergonzoni** pelo auxílio, conversas e amizade durante essa caminhada. Estendo os meus agradecimentos às alunas de iniciação científica **Dayanne Rodrigue e Júlia Barachini** pelo auxílio e companheirismo. Espero que, de alguma forma, eu tenha auxiliado na formação de vocês.

A minha namorada, amiga e companheira, **Roberta Abdala**. Obrigado pelo carinho, amor e compreensão. O meu sentimento por você, *loira*, representa o que de mais puro eu posso oferecer para alguém. Agradeço por toda motivação e companheirismo durante essa caminhada e por jamais me deixar fraquejar. Você, um dos meus maiores exemplos profissionais, faz parte dessa conquista. Amo você.

Ressalto o meu profundo agradecimento aos amigos que participaram dessa caminhada. Desde aqueles de longa data, vide **República Carrossel Caipira (C&C)** e **Handebol LAURP (FNT)**; aos que surgiram durante o percurso da pós-graduação, **Natália Agostinho (Bono)** e **Gabriel Favoretto (Brinquedo)** e **Carlos Kalva (Beisso)**; e também aos mais recentes, **Steneo, Carolina, Flávia, Henrique, Jéssica, Edivaldo e Beatriz**. Vocês, amigos, confortaram o percurso e me proporcionaram auxílio sempre que necessário. Obrigado.

Destaco o esporte, mas dentre as diversas modalidades que tive a oportunidade de vivenciar, ressalto o **Handebol**, o qual me trouxe amigos, empregos, aprendizados e diversas outras vivências marcantes. Obrigado por todas as equipes que tive oportunidade de trabalhar e/ou

jogar. Ademais, agradeço à todas as atletas e comissão técnica pela simpatia e paciência durante as coletas. Em especial, agradeço as atletas com quem tive a oportunidade de dividir o meu conhecimento.

Ao decorrer da minha caminhada acadêmica, fui agraciado com excelentes profissionais que me proporcionaram além de conhecimento, inspiração. Agradeço a todos os docentes da **Escola de Educação Física e Esporte de Ribeirão Preto**. Também agradeço aos funcionários e professores do **Programa de Pós-Graduação em Reabilitação e Desempenho Funcional da Faculdade de Medicina de Ribeirão Preto – Universidade de São Paulo**.

I would like to thank the Prof. Dra. Alice Mantoan and Dra. Monica Reggiane for the patience and the partnership during the BOPS process. Grazie.

Aos professores **Sérgio Augusto Cunha, Felipe Arruda Moura, Guilherme Manna César, Adelino Sanchez Ramos da Silva e Matheus Gomes Machado** que sempre me incentivaram a seguir em frente. Vocês são grandes exemplos de pesquisadores e professores.

O presente trabalho foi realizado com apoio da **Coordenação de Aperfeiçoamento de Pessoal de Nível Superior - Brasil (CAPES) - Código de Financiamento 001**. Também Agradeço à **CAPES** pela bolsa do **Programa de Doutorado Sanduíche no Exterior (PDSE) (#88881.189671/2018-01)**. Além disso, agradeço às demais instituições de fomento que proporcionaram a minha dedicação interina à pesquisa, podendo assim retribuir de alguma forma o que em mim foi investido: **Fundação de Amparo à Pesquisa do Estado de São Paulo (FAPESP #2016/09653-5; #2018/20362-8); Sociedade Brasileira de Biomecânica e Sociedade Internacional de Biomecânica** pelos prêmios concedidos a mim no decorrer dos últimos quatro anos.

EPÍGRAFE

Foi o tempo que dedicastes à tua rosa que a fez tão importante.

Antoine de Saint-Exupéry

RESUMO

BEDO, B. L. S. **Aplicação de um modelo musculoesquelético para análise biomecânica da articulação do joelho: efeito agudo da fadiga em atletas de handebol feminino**. 2021. 203 f. Tese (Doutorado) – Faculdade de Medicina de Ribeirão Preto, Universidade de São Paulo, Ribeirão Preto, 2021..

A presente obra apresenta, de forma crítica, uma seleção de trabalhos publicados e/ou em processo de revisão em revistas internacionais no campo da biomecânica, controle motor e bioengenharia. De modo geral, o conjunto das obras apresenta os possíveis efeitos da fadiga no controle postural e biomecânica dos membros inferiores em atletas de handebol do sexo feminino. Além disso, o documento apresenta ferramentas tanto para indução da fadiga quanto para análise biomecânica por meio do método de modelagem e simulação computacional em *OpenSim*. Os estudos que compõem o presente documento foram impulsionados por necessidades emergentes no meio acadêmico que, ao decorrer do processo do doutorado, foram percebidos tentando ser solucionadas. Um conjunto de cinco estudos empíricos buscam, de maneira cronológica, apresentar (i) um método para indução à fadiga; (ii) os efeitos da fadiga na cinemática dos membros inferiores; (iii) um modelo musculoesquelético para análise biomecânica da articulação do joelho, (iv) uma ferramenta para otimizar o processamento de dados de modelagem computacional e, por fim, (v) os efeitos de um protocolo de fadiga na biomecânica da articulação do joelho a partir da ótica de modelagem computacional. Ao todo, o texto fornece reflexões e possibilidades para contribuições gerais para o campo da biomecânica. Por fim, as limitações de cada uma das investigações apresentadas são discutidas no decorrer dos respectivos trabalhos, possibilitando direcionamentos para futuras investigações.

Palavras-chave: biomecânica; fadiga; modelagem computacional; *OpenSim*; handebol

ABSTRACT

BEDO, B. L. S. **Application of a musculoskeletal model to knee joint biomechanical analysis: acute effect of fatigue in female handball athletes.** 2021. 203 p. Tese (Doutorado) – Faculdade de Medicina de Ribeirão Preto, Universidade de São Paulo, Ribeirão Preto, 2021.

This current thesis presents, in a critical perspective, a selection of peer review and/or polished research articles in international journals of biomechanics, motor control and bioengineering fields. In general, the articles present the possible effect of fatigue on posture control and lower limb biomechanics in females handball athletes. Moreover, this text also presents tools for fatigue induction and for biomechanical analysis using the computational modeling and simulation method in OpenSim. The articles were shaped by the academic necessity, that during the doctoral process, were perceived and tried to be understood and remedied. A total of five empirical articles presents (i) a specific method to induce fatigue in handball athletes; (ii) the effect of fatigue on lower limb kinematics; (iii) a musculoskeletal model to knee joint analysis; (iv) a toolbox to optimize modeling and simulation analysis; and lastly (v) the impact of the fatigue in the knee biomechanics: a modelling approach. The text provides a critical reflection on the general contribution to the current body of scientific knowledge. Finally, the limitations of each study are discussed in each article, allowing a direction for future research.

Key Words: biomechanics; fatigue; computational modelling; OpenSim; handball

LISTA DE TABELAS

Capítulo 2 – Estudo 1

Table 1. Mean and standard deviation ($\pm$) of the fatigue parameters	55
---	----

Capítulo 4 – Estudo 3

Supplement Table 1. Joint range of motion and moment between the original and updated MSKM during gait and squat tasks. The results consider the left side as the side of interest.....	10
--	----

4

Supplement Table 2. Peak muscle forces and the maximum difference (MD) during gait and deep squat tasks. The results consider the left side as the side of interest.....	105
---	-----

Capítulo 5 – Estudo 4

Table 1. Details on open-source toolboxes in Figure 1 (colors on the left column refer to the corresponding step and toolbox main goal depicted in Figure 1). The used platform, the availability of Test Data, and the License are compared. Links to the project pages, software repositories, and available documentation are included	113
--	-----

Table 2. Summary of BOPS GUI functionalities. Detailed description of the GUI's main panels and functionalities available through the GUI graphical elements. The Matlab function styles are specified, as well as options available for the processing and any additional related information	117
---	-----

Table A1. Main functionalities of Open Source toolboxes based on OpenSim for musculoskeletal modeling and simulations	129
--	-----

Capítulo 6 – Estudo 5

Table 1. Mean and standard deviation ($\pm$) of the fatigue parameters	138
---	-----

LISTA DE FIGURAS

Visão geral da tese

Figura 1. Estrutura geral da tese que será apresentada no modelo escandinavo, composta por cinco artigos que, quando combinados, compõem o presente documento. Estudo 1: [Capítulo 2](#); Estudo 2: [Capítulo 3](#); Estudo 3: [Capítulo 4](#); Estudo 4: [Capítulo 5](#); Estudo 5: [Capítulo 6](#).9

Contextualização

Figura 1. Exemplo de uma mudança de direção realizada em ambiente laboratorial.12

Figura 2. Interface gráfica do OpenSim 3.3. Ao centro é possível observar um modelo musculoesquelético. À esquerda encontram-se os corpos, músculos, articulações, dentre outras ferramentas, que compõem o modelo. À direita encontra-se os valores angulares de cada uma das articulações que compõem o modelo com os respectivos valores.16

Figura 3. Organização do OpenSim e hierarquia estrutural (adaptado de Seth, Hicks, Uchida, Habib, and Dembia (2018) Fonte: https://simtk.org/api_docs/opensim/api_docs/).17

Figura 4. Típica sequência de elementos necessários para realização do movimento humano. O movimento é orquestrado pelos sistemas neural, muscular, esquelético e sensorial. Deste modo, a possibilidade de criar modelos computacionais para cada um desses sistemas, possibilita o estudo do movimento humano (Adaptado de Uchida and Delp (2020)).18

Figura 5. Elementos para análise inversa. Esse processo inicia-se com a utilização de um modelo biomecânico de escolha. Utilizando-se dos marcadores calculam-se os ângulos articulares, assim como a velocidade e aceleração dos segmentos. Em sequência, com utilização dos dados de força, obtêm-se os momentos articulares e, por fim, utilizando algum processo de otimização, calcula-se a força muscular e a força de reação articular (Adaptado de Uchida and Delp (2020)).18

Figura 6. Ilustração de um modelo musculoesquelético genérico (esquerda) e um com o processo de *scaling* finalizado (direita). O modelo à direita possui as características do sujeito e possibilita as análises disponíveis no modelo genérico. Fonte: OpenSim.20

Figura 7. Fluxograma da ferramenta de Dinâmica Inversa no OpenSim. As caixas verdes representam os dados experimentais; o modelo escolhido é representado pela caixa laranja; o arquivo de configuração da ferramenta é indicado pela cor azul e as caixas roxas representam arquivos gerados por análises disponíveis OpenSim.22

Figura 8. Fluxograma da ferramenta de Otimização Estática do OpenSim. As caixas verdes representam os dados experimentais; o modelo escolhido é representado pela caixa laranja; o arquivo de configuração da ferramenta de otimização é indicado pela cor azul e as caixas roxas representam arquivos gerados por análises disponíveis OpenSim.24

Figura 9. As linhas representam a força de reação, normalizadas pela massa do participante, na articulação tibiofemoral. As linhas de cor preta e vermelha representam os valores de força de reação articular menores e maiores que 2BW, respectivamente. Nesse caso, a força é aplicada e orientada no segmento distal (tíbia).25

Figura 10. Fluxograma das etapas necessárias para obtenção das variáveis presentes neste documento.26

Desafios metodológicos

Figura 1. Conversor Analógico-Digital (National Instruments #USB-6251-BNC, Austin, TX, USA) utilizado para aquisição e conversão dos sinais de força e momento da plataforma de força.35

Figura 2. Ilustração do salto contra movimento (*countermovement jump*) realizado pelas atletas para quantificar a máxima força de propulsão.36

Figura 3. Exemplo de um sinal de força durante a execução de um salto contra movimento (*countermovement jump*). O ponto vermelho representa a força máxima de propulsão antes do salto, enquanto o ponto azul representa a força pico de reação no solo.36

Figura 4. O gráfico da esquerda representa a força vertical de reação articular de um modelo que quantifica a força em um único ponto (preto) e a soma das componentes laterais/mediais (vermelho tracejado). Os gráficos da direita representam a força vertical no componente medial (topo) e lateral (abaixo) durante um agachamento para o modelo desenvolvido.38

Figura 5. Valores de abdução e adução da articulação do joelho durante a fase de contato na mudança de direção (cima). Nesse gráfico, as linhas pretas e vermelhas representam os valores de média e desvio padrão das condições 1 e 2, respectivamente. O ciclo investigado (ilustração dos membros inferiores) e os resultados da análise pareada realizada por meio do teste-t ($SPM\{t\}$) são apresentados na figura abaixo. As linhas tracejadas e números vermelhos representam os limites críticos para ocorrência de diferença significativa ($t^*=3.33$). Por fim, a área sombreada representa o período em que houve diferença significativa que, no caso, foi entre 0 e 10% do ciclo investigado com um valor de significância de $p=0.03$42

Capítulo 2 – Estudo 1

Figure 1. Illustration of the fatigue protocol based on specific movements performed by handball athletes. The athletes should perform the following actions in all of the stations as fast as possible: (A) Acceleration and sidestep cutting maneuver for the right and run until the end of the 3-m distance; receive a pass (at the end of the 3-m length) from a partner positioned 2.5 meters away (Partner 1); backshift until the 90°; run 3 m for the left side, and perform an accurate throw to a partner positioned 2.5 meters away (Partner 2); this task was performed twice per lap; (B) Ten double-leg lateral jumping/landing over a barrier (20-cm high), performed as fast as possible; (C) Sprint; (D) Double-leg plyometric jumps, performed as fast as possible; (E) Lateral shifting; (F) Sprint; (G) Back Shifting; and (H) Sprint with a 90° change

of direction. The protocol was divided into rounds, where the round number represents the number of laps in the circuit (i.e., round I [1 lap]; round II [2 laps]; round III [3 laps]; and so on) until the participant is unable to continue performing the protocol.53

Figure 2. Mean and standard error of the mean for the heart rate (top), rate of perceived exertion (middle) and percentage of vertical force at the countermovement jump (bottom) at the baseline and every minute after the fatigue protocol. $*p < 0.05$ $**p < 0.0001$57

Figure 3. Mean and standard error of the center of pressure (COP) mean for the area (top) and the root mean square in the anteroposterior (middle) and mediolateral (bottom) directions at the baseline (Base.) and every minute after the fatigue protocol. $*p < 0.05$ $**p < 0.0001$58

Capítulo 3 – Estudio 2

Figure 1. The schematic representation of the data collection. For the recording of the maximum force propulsion, a force plate was used during the countermovement jump (A and D). The participants performed three valid trials of each task before (B) and after (E) the fatigue protocol, which was based on motor skills performed during handball practice (C). The yellow leg indicates the limb evaluated.75

Figure 2. Hip (top) and knee (bottom) joint angles during the single-leg landing (SLL) task. The left, middle, and right plots represent the sagittal, frontal, and transversal planes, respectively. The red and black lines represent the average angles for the fatigue and baseline conditions, respectively. The corresponding colors of the shaded areas represent the standard deviation of each condition. The red bars with black mosaicking denote the significant differences areas indicated by the SPM analysis. The illustration in the bottom represents the time-normalized movement window investigated during the SLL task. The yellow leg represents the assessed limb.79

Figure 3. Hip (top) and knee (bottom) joint angles during the sidestep cutting maneuver (SCM) task. The left, middle, and right plots represent the sagittal, frontal, and transversal planes, respectively. The red and black lines represent the average angles for the fatigue and baseline conditions, respectively. The corresponding colors of the shaded areas represent the standard deviation of each condition. The red bars with black mosaicking denote the significant differences areas indicated by the SPM analysis. The illustration in the bottom represents the time-normalized movement window investigated during the SCM task. The yellow leg represents the assessed limb.80

Figure 4. Hip (top) and knee (bottom) joint angles during the drop vertical jump (DVJ) task. The left, middle, and right plots represent the sagittal, frontal, and transversal planes, respectively. The red and black lines represent the average angles for the fatigue and baseline conditions, respectively. The corresponding colors of the shaded areas represent the standard deviation of each condition. The red bars with black mosaicking denote the significant differences areas indicated by the SPM analysis. The illustration in the bottom represents the

time-normalized movement window investigated during the DVJ task. The yellow leg represents the assessed limb.81

Capítulo 4 – Estudo 3

Figure 1. The schematic musculoskeletal models (MSKM) that have been used in lower limb research and the implementation in the updated model. (A) Allows quantifying medial and lateral tibiofemoral contact forces. (B) Implemented musculotendon unit (MTU) parameters based on healthy and young individuals. (C) Increased knee range of motion and updated knee MTU parameters and path. (D) Controlled hip MTU moment arms to allow higher hip and knee flexion. The updated model (E) implemented the medial and lateral tibiofemoral contact forces strategy in an MSKM based on young and healthy MTU parameters and that allows a high range of motion of the hip and knee.95

Figure 2. Tibiofemoral contact forces (TFCF) during the gait (top) and squat (bottom) cycles for the original (blue) and updated (red) musculoskeletal model. The graphs on the left column represent the medial (top) and lateral (bottom) components reaction forces during the gait (A) and squat (B) cycles for the updated model. The graphs in the middle show the joint contact impulse calculated by the integral of TFCF from de original and updated models during the gait (C) and squat (D). The dark and light red columns represent the medial and lateral components, respectively. Lastly, the graphs on the right column show the body weight (BW) normalized TFCF resultant during the gait (E) squat (F) cycle, reported with the compass graph (posterior view). The blue lines show the magnitude (radar graph) and direction to the tibial plateau obtained from the original model. The red arrows represent the inverse of the peak TFCF of the lateral (light red) and medial (dark red) components obtained from the updated model.100

Supplement Figure 1. The topology view exported from OpenSim3.3® of the updated model. The right and left knee joints are composed by the medial (*med_cond_joint_r* and *med_cond_joint_l*) and lateral (*lat_cond_joint_r* and *med_cond_joint_l*) joints that share all tibiofemoral load on both components. The sagittal knee angles are based on the knee joint (*knee_r* and *knee_l*) that are connected between the parent (*femoral_cond_r* and *femoral_cond_l*) and the child (*sagittal_articulation_frame_r* and *sagittal_articulation_frame_l*).106

Capítulo 5 – Estudo 4

Figure 1. Workflow for neuro and musculoskeletal modeling applications and list of toolboxes available to accomplish each step. Open-source toolboxes are highlighted in bold.111

Figure 2. BOPS GUI. Layout of the user-friendly graphical Matlab interface available in BOPS. It allows to select which OpenSim procedures to perform sequentially, among *Inverse Kinematics*, *Inverse Dynamic*, *Muscle Analysis*, *Static Optimization*, and *Joint Reaction Analysis*, and to input both processing parameters and corresponding setup files. Please, refer to Table 2 for a detailed description of the GUI functionalities.116

Figure 3. BOPS folders organization. The structure on the left presents the data organization suggested by the authors. All input data must be grouped in a *dynamicElaborations* folder, and subdivided in trials' folders that include both *.mot* and *.trc* files. This file organization can be easily obtained using MOTO NMS software. The OpenSim model file (*.osim*) can be automatically loaded on the BOPS GUI if located in a dedicated folder as recommended. The output data folders include processing results, figures, logs, and setup files and are automatically created mirroring the structure of the *dynamicElaborations* folder. On the right, BOPS main folder organization is presented. The *.src* folder has the main source code while the *Templates* folder stores the setup files necessary to run the scripts.118

Figure 4. Testing Procedure Results. Examples of inverse kinematic, inverse dynamic, static optimization and joint reaction results from BOPS (red) and OpenSim (black) during the stance phase of a gait trial. The results obtained with the two approaches fully overlap.122

Capítulo 6 – Estudio 5

Figure 1. Knee flexion angle during sidestep cutting maneuver. The black and red lines represent the average angles for the baseline and fatigue conditions, respectively. The corresponding colors of the shaded areas represent the standard deviation of each condition. The black bars with red striped denote the significant differences areas indicated by the SPM analysis.139

Figure 2. Vastus muscles forces during sidestep cutting maneuver task. The red and black lines represent the average angles for the fatigue and baseline conditions, respectively. The black and red lines represent the average muscle forces for the baseline and fatigue conditions, respectively. The corresponding colors of the shaded areas represent the standard deviation of each condition. The black bars with red striped denote the significant differences areas indicated by the SPM analysis.140

Figure 3. Tibiofemoral contact forces on anterior-posterior (A), superior-inferior (B) and medial-lateral (C) directions during sidestep cutting maneuver task. The red and black lines represent the average of tibiofemoral forces for the fatigue and baseline conditions, respectively. The corresponding colors of the shaded areas represent the standard deviation of each condition. The black bars with red striped denote the significant differences areas indicated by the SPM analysis.140

Figure 4. Tibiofemoral contact forces during the sidestep cutting maneuver. The graphs on the left column represent the medial (A) and lateral (B) components' reaction forces during the task. The red and black lines represent the average of tibiofemoral forces for the fatigue and baseline conditions, respectively. Lastly, the graphs on the right column show the body weight (BW) normalized tibiofemoral contact forces resultant during the task, reported with the compass graph on lateral (C) and posterior view (D). The red and black lines show the magnitude (radar graph) and direction to the tibial plateau obtained from the musculoskeletal model.141

LISTA DE SIGLAS E ABREVIATURAS

ACL – <i>Anterior Cruciate Ligament</i>	LSD – <i>Least Significant Difference</i>
API – <i>Application Programming Interface</i>	MAP – <i>Musculoskeletal Atlas Project</i>
BOPS – <i>Batch OpenSim Processing Scripts</i>	MMII – <i>Membros Inferiores</i>
BW – <i>Body Weight</i>	MSKM – <i>Musculoskeletal models</i>
CAPEs – Coordenação de Aperfeiçoamento de Pessoal de Nível Superior	MTU – <i>Musculotendon unit</i>
CIHR – <i>Canadian Institutes of Health Research</i>	MS/s – <i>Million Samples per Second</i>
COP – <i>Center of Pressure</i>	M_x – Momento de Força no eixo x
DVJ – <i>Drop Vertical Jump</i>	M_{xd}, M_{yd} e M_{zd} – Momentos distais em x, y e z
EMG-driven – <i>electromyography-driven</i>	M_y – Momento de Força no eixo y
EQ – Erro Quadrado	M_z – Momento de Força no eixo z
FAPESP – Fundação de Amparo à Pesquisa do Estado de São Paulo	RMS – <i>Root Mean Square</i>
F_{MAX} – Máxima Força de Propulsão	RPE – <i>Rate of Perceived Exertion</i>
F_x – Componente de Força no eixo x	RPE – <i>Rate of Perceived Exertion</i>
F_y – Componente de Força no eixo y	R_{xp}, R_{xd}, R_{yp}, R_{yd}, R_{zp} e R_{zd} – Forças de reação das articulações nos eixos x, y e z
F_z – Componente de Força no eixo z	SCM – <i>Sidestep Cutting Maneuver / Mudança de Direção</i>
GUI – <i>Graphical User Interface</i>	SLL – <i>Aterrissagem Unipodal / Single Leg Landing</i>
HR – <i>Heart Rate</i>	SO – <i>Static Optimazation</i>
HR – <i>Reart Rate</i>	SPM – <i>Statistical Parametric Mapping</i>
ID – <i>Inverse Dynamic</i>	TFCF – <i>Tibiofemoral Contact Forces</i>
IK – <i>Inverse Kinematic</i>	UOMAM – <i>University of Ottawa Motion Analysis Model</i>
I_x I_y e I_z – Momentos de inércia em x, y e z	α_x, α_y e α_z – Componente da aceleração angular em x, y e z
JRA – <i>Joint Reaction Analysis</i>	ω_x, ω_y e ω_z – Componente da velocidade angular em x, y e z
LCA – <i>Ligamento Cruzado Anterior</i>	
l_p, l_d – Distância do centro de massa das articulações distal e proximal.	

SUMÁRIO

Dedicatória.....	5
Agradecimentos	6
Epígrafe.....	10
Resumo	11
Abstract.....	12
Lista de tabelas.....	13
Capítulo 2 – Estudo 1	13
Capítulo 4 – Estudo 3	13
Capítulo 5 – Estudo 4	13
Capítulo 6 – Estudo 5	13
Lista de figuras.....	14
Visão geral da tesa.....	14
Contextualização	14
Desafios metodológicos.....	15
Capítulo 2 – Estudo 1	15
Capítulo 3 – Estudo 2	16
Capítulo 4 – Estudo 3	17
Capítulo 5 – Estudo 4	17
Capítulo 6 – Estudo 5	18
Lista de siglas e abreviaturas	19
Capítulo 1.....	4
2.1 Visão geral da tese.....	5
2.1.1 Estrutura.....	5
2.2 Contextualização	10
2.2.1 Incidência e mecanismo de lesão na articulação do joelho	10
2.2.2 Modelagem e simulação do sistema musculoesquelético.....	14
2.3 Objetivo geral.....	31
2.4 Estudos Empíricos.....	31
2.5 Desafios metodológicos	33
Protocolo de fadiga.....	33
Quantificação da força durante o protocolo	34
Desenvolvimento de um modelo musculoesquelético e de uma ferramenta de otimização de análise de dados em OpenSim	37
Statistical parametric mapping	39
Capítulo 2.....	46

Estudo 1.....	46
Capítulo 3.....	69
Estudo 2.....	69
Capítulo 4.....	91
Estudo 3.....	91
Capítulo 5.....	107
Estudo 4.....	107
Capítulo 6.....	130
Estudo 5.....	130
Capítulo 7.....	155
Discussão Geral e Considerações finais.....	155
Capítulo 9.....	159
Apêndices.....	159
Apêndice 1 – Ficha de avaliação.....	160
Apêndice 2 – Ficha de Coleta	163
Apêndice 3 – Aprovação Comitê de Ética	165
Apêndice 4 – Termo de Consentimento Livre e Esclarecido.....	167
Capítulo 10.....	170
ANEXOS.....	170
Anexo 1 – Script para quantificação dos dados de força	171
Anexo 2 – Estudo 1	174
Anexo 3 – Estudo 2	175
Anexo 4 – Estudo 3	176
Anexo 5 – Estudo 4	177

CAPÍTULO 1

VISÃO GERAL DA TESE

Estrutura | Contextualização | Estudos Empíricos | Desafios Metodológicos

2.1 VISÃO GERAL DA TESE

2.1.1 Estrutura

A presente tese foi composta no modelo escandinavo, constituída pelo conjunto de artigos submetidos, aceitos ou publicados em revistas internacionais na área de biomecânica e bioengenharia. Embora não diretamente conectas, as obras apresentam cronologicamente a evolução do presente projeto orientado para elucidar o efeito de um fator intrínseco (fadiga) na biomecânica da articulação do joelho. Além desse questionamento, as obras apresentam novos métodos e técnicas que auxiliarão futuros pesquisadores. Ao todo, o presente documento é composto por cinco estudos empíricos que foram cuidadosamente elaborados.

Dado que a fadiga é um dos pontos centrais da presente tese, foi-se necessário desenvolver um método para sua indução. Buscando uma validade externa ecológica, ou seja, um método que simulasse ações realizadas na modalidade investigada (handebol) e que permitisse a realização em um ambiente laboratorial, pensou-se em um protocolo de fadiga em formato de circuito, composto por uma sequência de ações realizadas constantemente por atletas de handebol na prática esportiva. Visando os artigos sequenciais da tese, viu-se necessário quantificar o tempo de duração da fadiga induzida para que futuras investigações para averiguar o efeito da fadiga durante a realização de determinadas tarefas realizassem as mesmas no período de permanência da fadiga. Partindo desses questionamentos, originou-se o Estudo 1 ([Capítulo 2](#) – páginas amarelas) intitulado “*The rapid recovery of vertical force propulsion production and postural sway after a specific fatigue protocol in female handball athletes*”.

Com a compreensão do tempo de duração da fadiga, realizou-se o Estudo 2 ([Capítulo 3](#) – páginas azuis). Neste estudo, objetivamo-nos em averiguar o efeito da fadiga, oriunda de um protocolo baseado em ações realizadas na prática do handebol, na cinemática da articulação

do joelho na execução de mudanças de direção e aterrissagens unipodais. Baseado nesses questionamentos, originou-se o artigo “*The effect of fatigue on hip and knee kinematics during single and double-leg dynamic tasks: A study on female handball players*”. Após observar e discutir os resultados encontrados nesse artigo, nos encontramos em um impasse. Embora os resultados indicassem que a fadiga poderia afetar a carga interna nas articulações dos membros inferiores, não possuíamos embasamentos científicos para tal afirmação. Assim, encontramos a necessidade de nos direcionar para análises biomecânica mais robustas e detalhadas.

Visando solucionar tal lacuna, passei a interessar-me na compreensão de técnicas de modelagem e simulação computacional. Entre as possibilidades de análises que a modelagem e simulação computacional possibilitam, encontra-se o cálculo de força de reação articular de forma não invasiva. Entretanto, a maioria dos modelos musculoesqueléticos com foco na articulação do joelho baseava-se no cálculo de forças de reação interarticular em um único ponto da articulação tibiofemoral. Deste modo, o Estudo 3 ([Capítulo 4](#) – páginas verdes), intitulado “*A custom musculoskeletal model for estimation of medial and lateral tibiofemoral contact forces during tasks with high knee and hip flexions*” apresenta e valida um modelo musculoesquelético para cálculo de forças de reação articular nas porções mediais e laterais na articulação tibiofemoral. Ademais, o modelo permite simulações computacionais de tarefas com expressivos valores de flexão de joelho e quadril. O desenvolvimento deste modelo abriu uma nova possibilidade para responder o questionamento central desta tese.

Apesar das análises de modelagem e simulação computacional possibilitarem uma nova perspectiva no que concerne a avaliações biomecânicas, tais métodos são computacionalmente custosos do ponto de vista de processamento, além de demandarem conhecimento profundo da técnica. Além disso, durante a etapa de processamento de dados de simulação e/ou modelagem computacional, são necessárias diversas etapas metódicas que, casualmente, podem resultar em erros.

Para otimizar o processo de análise e minimizar a possibilidade de erros, elaborou-se o Estudo 4 ([Capítulo 5](#) – páginas laranjas) intitulado “*BOPS: A Matlab toolbox to Batch Musculoskeletal Data Processing from OpenSim*”, que apresenta um *toolbox* que possibilita a otimização de tempo durante o processo de análises de modelagem e simulação computacional. Além disso, a ferramenta permite que profissionais que não possuam vasta experiência em programação e/ou modelagem computacional, se favoreçam das possibilidades dessas análises.

Após desenvolver métodos e técnicas biomecânicas para responder o questionamento central da presente tese, nos encontramos na possibilidade averiguar como a fadiga, induzida por um protocolo composto por ações frequentemente realizadas durante a prática do handebol, afetaria os membros inferiores de atletas do sexo feminino. Assim, o Estudo 5 ([Capítulo 6](#) – páginas roxas) “*Effect of the fatigue in knee biomechanics during the sidestep cutting maneuver: a modeling approach.*” foi desenvolvido para analisar o efeito da fadiga na biomecânica, a partir da ótica da modelagem computacional, na articulação do joelho na execução da mudança de direção.

Para responder o questionamento central da presente tese, foram utilizadas todas as técnicas e métodos apresentados nos estudos anteriores. Para induzir a fadiga, utilizou-se o protocolo de ações do handebol apresentado no [Estudo 1](#). Pela tarefa realizada (mudança de direção) requisitar expressivos valores de flexão de joelho e quadril ([Estudo 3](#)), como a necessidade de calcular as forças de reação articular nos côndilos mediais e laterais da articulação do joelho, utilizou-se o modelo proposto no [Estudo 2](#). Para otimizar o processamento de dados e minimizar os erros no processamento de dados, utilizou-se o *toolbox* apresentado no [Estudo 4](#) para realizar as análises de modelagem computacional (Figura 1).

Julgo importante ressaltar que os estudos apresentados refletem a caminhada acadêmica ao longo dos últimos quatro anos (2017 a 2020). Embora alguns dos trabalhos não compunham o projeto inicial, a problemática resultante da minha interação com diversos pesquisadores do

campo da biomecânica, culminou no planejamento e execução dos estudos descritos e, quando concatenados, resultaram na presente tese.

Figura 1. Estrutura geral da tese que será apresentada no modelo escandinavo, composta por cinco artigos que, quando combinados, compõem o presente documento. Estudo 1: [Capítulo 2](#); Estudo 2: [Capítulo 3](#); Estudo 3: [Capítulo 4](#); Estudo 4: [Capítulo 5](#); Estudo 5: [Capítulo 6](#).

2.2 CONTEXTUALIZAÇÃO

2.2.1 Incidência e mecanismo de lesão na articulação do joelho

O termo lesão refere-se a qualquer dano que acometa tecidos moles e rígidos do corpo humano (Whiting e Zernicke 2009). As lesões esportivas remetem-se a quaisquer lesões oriundas de atividades e exercícios físicos (Hoff e Martin 1986, van Mechelen, Hlobil et al. 1992, Caine, Caine et al. 1996). Embora em alguns contextos as lesões possam ser prevenidas, necessitam-se estudos longos e complexos para que, quando culminados, possibilitem a redução da incidência de uma determinada lesão. Para tal processo, van Mechelen, Hlobil et al. (1992) descreveram uma sequência de etapas científicas que devem ser realizadas visando à prevenção. Dentre os processos sugeridos, mais especificamente a primeira etapa, encontra-se a classificação e severidade das lesões que acometem os atletas de um determinado esporte.

Partindo desse pressuposto, como a maioria das modalidades coletivas de esportes de invasão (futebol, futsal, basquete, rugby, por exemplo), o handebol requisita de seus atletas frequentes mudanças de direção, saltos, aterrissagens com e sem contato com adversários. O conjunto dessas características com a superfície de prática (antiaderente) tornam o handebol uma das modalidades com maior incidência de lesões nos membros inferiores (MMII) (Noyes, Dunworth et al. 1996, Mandelbaum, Silvers et al. 2005, Olsen, Myklebust et al. 2006) , sobretudo nas articulações do joelho e do tornozelo (Wedderkopp, Kaltoft et al. 1997, Wedderkopp, Kaltoft et al. 1999, Olsen, Myklebust et al. 2004, Olsen, Myklebust et al. 2006, Cumps, Verhagen et al. 2007, Sanches e Borin 2009, Bedo, Manechini et al. 2019), na qual a ruptura do ligamento cruzado anterior (LCA) é uma das lesões mais frequentes (Hewett, Myer et al. 2001, Olsen, Myklebust et al. 2006, Cochrane, Lloyd et al. 2007).

No que concerne aos indivíduos mais acometida por rupturas do LCA, devido a fatores hormonais (relação entre picos de hormônios sexuais femininos e frouxidão ligamentar) (EA

Arendt, Bershadsky, & Agel, 2002), biomecânicos (ângulo Q) (Hewett, Myer et al. 2005), anatômicos (espessura do LCA) (Griffin, Agel et al. 2000, Griffin, Albohm et al. 2006) e neuromusculares (diferenças nos padrões de ativação quadríceps em relação aos isquiotibiais) (Hewett 2000, Hewett, Paterno et al. 2002, Hewett, Myer et al. 2005), a população do sexo feminino encontra-se mais exposta à ocorrência dessa lesão. Deste modo, a conciliação desses fatores intrínsecos e a modalidade praticada, tornam atletas de handebol do sexo feminino uma população com expressivo risco de lesão na articulação do joelho (Lindblad, Hoy et al. 1992, Wedderkopp, Kaltoft et al. 1997, Hewett, Ford et al. 2005, Langevoort, Myklebust et al. 2007, Moller, Attermann et al. 2012, Bedo, Manecchini et al. 2019).

Uma vez compreendido o aspecto epidemiológico, dá-se sequência ao fluxograma apresentado por W. van Mechelen et al. (1992) para, possivelmente, obter a prevenção de uma lesão. A segunda etapa remete-se à compreensão da etiologia e do mecanismo das principais lesões encontradas em uma determinada modalidade. De modo resumido, no que concerne à ruptura do LCA, sabe-se que na maioria dos casos (cerca de 70 a 85%) as lesões acontecem sem contato direto (Arendt e Dick 1995, Griffin, Agel et al. 2000, Alentorn-Geli, Myer et al. 2009, Micheo, Hernandez et al. 2010), durante movimentos com desacelerações (Olsen, Myklebust et al. 2004); aterrissagens (Zebis, Andersen et al. 2009) e atividades de pivô, com a articulação próxima a sua extensão ou, da mesma forma, com baixo valores de flexão do joelho.

Assim, diferentes tarefas são realizadas em ambientes laboratoriais visando emular as ações realizadas durante a prática esportiva. Como exemplo de movimentos com maior risco de ocorrência de lesões no joelho, em especial a ruptura do LCA, encontram-se as mudanças de direção, em inglês, *sidestep cutting maneuver* e aterrissagens realizadas de modo bipodal ou unipodal (Figura 1). No que tange à mudança de direção, esse movimento é requisitado, em sua maioria, visando ludibriar um adversário ou, no caso defensivo, na necessidade de uma mudança de direção repentina para interferir em uma ação ofensiva do adversário. Quanto aos

saltos, eles são observados em diferentes contextos, como cabeceios (futebol/futsal) e arremessos (handebol e basquetebol).

Figura 1. Exemplo de uma mudança de direção realizada em ambiente laboratorial.

Durante a realização dessas tarefas, a combinação de diferentes movimentos nas articulações dos membros inferiores e tronco resultam na possibilidade de ocorrência da lesão. Em suma, o aumento do risco de ocorrência da ruptura do LCA se deve a combinação da rotação externa da tibia próximo do instante em que há uma baixa flexão da articulação do joelho (20 a 30°) (Cochrane, Lloyd et al. 2007, Koga, Nakamae et al. 2010) e o movimento ou posicionamento da região distal da tibia para longe da linha sagital mediana do corpo, enquanto a região distal do fêmur está mais próxima da linha sagital mediana, aumenta o risco da ocorrência da ruptura do LCA (Hewett, Ford et al. 2005). Deste modo, torna-se fundamental a compreensão das cargas articulares durante a realização dessas tarefas.

Como descrito, a ocorrência de lesões depende de diversos fatores que permeiam entre intrínsecos e extrínsecos (Whiting e Zernicke 2009). Neste sentido, a fadiga, definida como a incapacidade do músculo produzir força devido a indução por um exercício (Gandevia 2001) ou como a incapacidade de realizar quaisquer funções físicas e/ou cognitivas devido a fadigabilidade de desempenho e a percebida (Enoka e Duchateau, 2016), pode possivelmente influenciar no risco da ocorrência de uma lesão, principalmente se tratando das interações física e cognitivas que a fadiga pode resultar. Essa afirmação baseia-se na teoria de que os músculos, quando fadigados, reduzem a sua capacidade de absorção de energia (Mair, Seaber et al. 1996).

Além disso, a fadiga também relaciona-se com o agravamento no controle motor, aumento da frouxidão ligamentar (Skinner, Wyatt et al. 1986, Wojtys, Beaulieu et al. 2016), agravo do equilíbrio (Johnston, Howard et al. 1998) e redução da propriocepção (Lattanzio, Petrella et al. 1997, Hiemstra, Lo et al. 2001, Allen e Proske 2006), mecanismos importantes para evitar a ocorrência de uma lesão. A propriocepção refere-se ao processo de aquisição e processamento dos estímulos sensoriais oriundos dos mecanorreceptores localizados nos músculos, tendões, ligamentos e pele, para manter a estabilidade dinâmica da articulação (Relph 2015). Entretanto, quando sua capacidade é diminuída, como em situações de fadiga, os atletas tornam-se mais vulneráveis a ocorrência de lesões. Como exemplo, estudos apontam que a maioria das lesões ocorre no final de cada tempo do jogo (Stuart e Smith 1995, Chappell, Herman et al. 2005).

Em contrapartida, embora diversos estudos apontem que a fadiga viabiliza alterações no sistema proprioceptivo que, ocasionalmente, resultam em alterações prejudiciais para as articulações, a afirmação de que a fadiga aumenta o risco de lesões ainda permanece desconhecida e não é inteiramente correta (Bourne, Webster et al. 2019). Como exemplo, apesar de alguns pesquisadores encontrarem que a fadiga muscular está relacionada com o aumento do risco de lesões do LCA (McLean, Fellin et al. 2007, Borotikar, Newcomer et al. 2008, Liederbach, Kremenec et al. 2014, Xia, Zhang et al. 2017), uma recente revisão de literatura sistemática apontou que os protocolos de fadiga publicados até o presente momento não afetaram variáveis cinemáticas e dinamométricas que relacionam-se diretamente com o aumento do risco de lesões (Benjaminse, Webster et al. 2019). Do mesmo modo, estudos epidemiológicos não apresentaram relação entre o tempo de permanência em quadra e fadiga com a ocorrência de lesão (Waldén, Häggglund et al. 2011). Contudo, ainda existem poucas investigações que quantificaram parâmetros adequados para estimar efeitos da fadiga e possíveis alterações que aumentem o risco de lesão no LCA (Bourne, Webster et al. 2019). Dentro desse contexto, torna-se necessário o desenvolvimento e utilização de abordagens

biomecânicas que possibilitem uma compreensão mais robusta no tocante a biomecânica articular como, por exemplo, a utilização de modelagem e simulação computacional mostra-se que são ferramentas importantes para compreensão das cargas articulares.

2.2.2 Modelagem e simulação do sistema musculoesquelético.

Como apresentado nos parágrafos anteriores, diversas técnicas têm sido utilizadas para uma melhor compreensão de alterações nas articulações. Para investigar interações complexas entre os sistemas neuromuscular e esquelético, técnicas invasivas que permitam cálculos de variáveis mais específicas como a força muscular e de reação articular (Arnold, Blemker et al. 2001, Delp, Anderson et al. 2007) são necessárias. Contudo, devido ao avanço tecnológico, com a utilização de alguns modelos computacionais e matemáticos, torna-se possível a estimação de variáveis internas de modo inteiramente não invasivo. Para isso, modelos de simulação *in silico* (computacional) otimizaram a compreensão do movimento e como os seus componentes interagem.

A modelagem do sistema neuromusculoesquelético viabiliza a compreensão do movimento e dos seus componentes de modo não invasivo. Para isso, diversos *softwares* foram desenvolvidos possibilitando esse tipo de análise como OpenSim (*Stanford University, USA*) (Delp, Anderson et al. 2007, Seth, Sherman et al. 2011, Seth, Hicks et al. 2018), SIMM (*Motion Analysis, USA*) (Delp, Loan et al. 1990) e AnyBody (*AnyBody Technology, Denmark*) (Damsgaard, Rasmussen et al. 2006). O OpenSim é uma escolha frequente entre pesquisadores da biomecânica. Desenvolvido na Universidade de Stanford, pelos pesquisadores Delp, Anderson et al. (2007) e Seth, Sherman et al. (2011) é uma plataforma de código aberto (*open source*) que possibilita a análises de simulação e modelagem computacional de um sistema neuromusculoesquelético. De forma resumida, essa ferramenta permite diferentes tipos de análises: (i) visualização de dados experimentais e de simulação;

(ii) simulação e construção de modelos musculoesqueléticos; e (iii) realização de diferentes tipos de análises biomecânicas, como cinemática e dinâmica inversa, cálculo de forças nos músculos e articulações, entre outras. Além disso, a ferramenta de escalonamento (em inglês, *scaling*) possibilita a criação de um modelo individual para cada participante (Seth, Sherman et al. 2011, Seth, Hicks et al. 2018)

Além das possibilidades descritas, o OpenSim permite a interação entre pesquisadores por meio de uma plataforma online. A plataforma possibilita o compartilhamento de bibliotecas, modelos musculoesqueléticos e tutoriais para testagem gratuitamente. Devido a isso, a comunidade do OpenSim possibilita e facilita a interação multicêntrica em projetos, além de potencializar o crescimento em grande escala de usuários da ferramenta.

O OpenSim¹, que tem a sua base desenvolvida em C++, oferece duas interfaces para os usuários. A primeira, e mais conhecida, é uma interface gráfica, desenvolvida em Java, que possibilita ao usuário a utilização da ferramenta sem a necessidade de maiores conhecimentos de programação (Figura 2). A segunda é por meio de uma API (sigla para o termo em inglês, *application programming interface*), que fornece aos pesquisadores uma forma de customizar e/ou criar modelos musculoesqueléticos, e processar dados de modelagem e/ou simulação utilizando linguagens como Matlab e Python. Com essa interface, torna-se possível a realização de cálculos de simulação de modelagem sem a utilização da interface gráfica do OpenSim, assim como a criação de ferramentas que otimizem o tempo de processamento (Maiores informações sobre a API serão apresentadas no [Capítulo 5](#)).

Contudo, antes de compreender quais os procedimentos necessários para a realização de análises biomecânicas utilizando modelagem e simulação computacional, é importante salientar quais são os procedimentos para obtenção de variáveis biomecânicas.

¹ OpenSim User's Guide: simtk-confluence.stanford.edu:8443/display/OpenSim33/User%27s+Guide
Scripting with MATLAB: <http://simtk-confluence.stanford.edu:8080/display/OpenSim/Scripting+with+Matlab>
API OpenSim Documentação: https://simtk.org/api_docs/opensim/api_docs/

Figura 2. Interface gráfica do OpenSim 3.3. Ao centro é possível observar um modelo musculoesquelético. À esquerda encontram-se os corpos, músculos, articulações, dentre outras ferramentas, que compõem o modelo. A direita encontra-se os valores angulares de cada uma das articulações que compõem o modelo com os respectivos valores.

Partindo de uma abordagem natural, para realização do movimento necessita-se de um padrão de ativação neural que contrai e relaxa os músculos (coativação muscular / co-contracção / inibição recíproca) e eles, por sua vez, resultam na locomoção (Figura 3). Primeiramente um comando central, seja ele voluntário ou involuntário, estimula as fibras musculares que, por sua vez, produzem torque nas articulações por meio dos tendões. Assim, com a força aplicada sobre uma superfície e sua reação no corpo humano, deixa-se a inércia e inicia-se o movimento.

Figura 3. Organização do OpenSim e hierarquia estrutural (adaptado de Seth, Hicks et al. (2018) Fonte: https://simtk.org/api_docs/opensim/api_docs/)

Entretanto, não é possível a realização de tais procedimentos de forma linear em análises biomecânicas. Assim, torna-se necessário a aplicação do modelo reverso (Figura 4), que é uma estratégia para combinar dados laboratoriais com modelos musculoesqueléticos previamente elaborados para o cálculo de diferentes variáveis e simular o movimento humano.

De forma resumida, primeiramente convertem-se as coordenadas dos marcadores retrorrefletivos em marcadores virtuais de um determinado modelo. Posteriormente, calculam-se os ângulos utilizando a cinemática inversa. Após a derivação dos valores angulares, junto com a ação de aceleração da gravidade e massa dos segmentos, calculam-se os momentos articulares por meio da técnica de dinâmica inversa. Em sequência, utiliza-se um método de otimização (estático ou dinâmico) para cálculo das forças musculares e, por fim, o cálculo das forças articulares. Esse processo é chamado de análise inversa (Uchida e Delp 2020). Maiores explicações sobre as etapas necessárias para esses procedimentos serão apresentadas nos tópicos seguintes.

Figura 4. Típica sequência de elementos necessários para realização do movimento humano. O movimento é orquestrado pelos sistemas neural, muscular, esquelético e sensorial. Deste modo, a possibilidade de criar modelos computacionais para cada um desses sistemas, possibilita o estudo do movimento humano. Adaptado de Uchida e Delp (2020).

Figura 5. Elementos para análise inversa. Esse processo inicia-se com a utilização de um modelo biomecânico de escolha. Utilizando-se dos marcadores calculam-se os ângulos articulares, assim como a velocidade e aceleração dos segmentos. Em sequência, com utilização dos dados de força, obtêm-se os momentos articulares e, por fim, utilizando algum processo de otimização, calcula-se a força muscular e a força de reação articular. Adaptado de Uchida e Delp (2020).

2.2.2.1 Escalonamento (Scaling)

Assim como as diversas possibilidades de análises e variáveis disponíveis por meio do OpenSim, proporcionalmente estão disponíveis modelos musculoesqueléticos para utilização (Delp, Anderson et al. 2007, Rajagopal, Dembia et al. 2016, Seth, Hicks et al. 2018, Catelli,

Wesseling et al. 2019, Bedo, Catelli et al. 2020). Cada modelo, com suas particularidades e foco em um determinado contexto de análise, são compartilhados de modo genérico. Ou seja, embora possuam suas características claramente descritas, quando utilizados, eles precisam ser adaptados para cada participante. Um exemplo, é o tamanho do modelo, geometrias dos corpos rígidos que compõem o modelo e sua massa. Assim, o processo de adaptação de um modelo genérico para cada participante é fundamental para a fidedignidade dos resultados.

Esse processo de adaptação de um modelo genérico para cada sujeito é denominado escalonamento (em inglês, *scaling*²), na qual o intuito é combinar os marcadores virtuais do modelo genérico com os marcadores obtidos por meio de uma tomada estática (Delp, Anderson et al. 2007), visando o desenvolvimento de um modelo musculoesquelético específico para aquele sujeito mas que, ao mesmo tempo, respeite os parâmetros estipulados no modelo genérico (Seth, Sherman et al. 2011) (Figura 5).

O sucesso na criação de um modelo individual para cada sujeito, mede-se pela comparação das distâncias dos marcadores originais (modelo genérico) com os marcadores experimentais (arquivos das coletas de cada participante). Essa adaptação é primeiramente feita de forma automática pelo OpenSim; contudo, uma alternativa em pequenos casos é adaptar marcadores manualmente para a sua adaptação com o modelo genérico. Entretanto, sugere-se fortemente que esse processo não seja realizado nas proeminências ósseas fundamentais para elaboração dos segmentos.

Ao final, utiliza-se o fator de *scaling* tamanho-dependente (em inglês, *length-dependent scaling factor*) (Delp, Anderson et al. 2007). Esse fator é calculado individualmente para cada marcador e, por isso, sugere-se que os valores sejam os menores possíveis, especialmente nos marcadores fundamentais para criação dos segmentos. Após esse processo, espera-se que os segmentos possuam tamanhos fidedignos aos originais do(a) participante.

² How scaling work: <https://simtk-confluence.stanford.edu/display/OpenSim/How+Scaling+Works>

Figura 6. Ilustração de um modelo musculoesquelético genérico (esquerda) e um com o processo de *scaling* finalizado (direita). O modelo à direita possui as características do sujeito e possibilita as análises disponíveis no modelo genérico. Fonte: OpenSim.

2.2.2.2 Cinemática Inversa (Inverse Kinematic)

Após elaborar o modelo individual para cada sujeito, o método da cinemática inversa possibilita encontrar o melhor conjunto de coordenadas que representem fidedignamente o realizado durante a aquisição dos dados. Para isso, o método possibilita identificar a posição dos marcadores, como os valores angulares de posição de cada articulação presente no modelo.

Dado que os modelos musculoesqueléticos possuem marcadores virtuais em proeminências ósseas previamente conhecidas, durante o processo de coleta dos dados, os marcadores são posicionados nas mesmas localidades anatômicas. Assim, durante o processo da cinemática inversa encontram-se as posições dos marcadores quadro a quadro e, por fim, utilizando os ângulos de Cardan (Cappozzo, Catani et al. 1995), obtém-se os ângulos articulares

Além disso, como descrito por Delp, Anderson et al. (2007), na cinemática inversa

realiza-se a minimização do erro entre os marcadores experimentais (coletados com o sistema de aquisição) e os virtuais. Matematicamente, esse processo é realizado pela seguinte equação:

$$EQ = \sum_{i=1}^{Marcadores} w_i (X_i^{\rightarrow experimental} - X_i^{\rightarrow virtual})^2 + \sum_{h=1}^{angulos} w_j (\theta_j^{experimental} - \theta_j^{virtual})^2$$

Na qual EQ é o erro quadrado, $X_i^{\rightarrow experimental}$ e $X_i^{\rightarrow virtual}$ são as posições tridimensionais dos marcadores do experimento e o virtual, i^{th} e $\theta_j^{experimental}$ e $\theta_j^{virtual}$ são os valores angulares da articulação j^{th} , sendo que w_i e w_j são os fatores que permitem os marcadores terem pesos diferenciados (Delp, Anderson et al. 2007). Esses pesos são estabelecidos pelo pesquisador e a sua prioridade ou “força” no momento da realização dos cálculos é proporcional ao valor indicado. Quanto maiores os pesos, maior é a prioridade que o programa oferecerá para ele no momento do cálculo. Portanto sugere-se que os marcadores com maior peso sejam aqueles posicionados nas proeminências ósseas fundamentais e, em contrapartida, os de menores pesos, aqueles posicionados no meio dos segmentos.

Ao fim desta etapa, o OpenSim fornece um arquivo contendo as coordenadas de cada marcador, quadro a quadro, assim como os valores angulares de cada articulação presente no modelo escolhido. Esses resultados são fundamentais para as análises seguintes.

2.2.2.3 Dinâmica Inversa (Inverse Dynamics)

Após obter as coordenadas dos marcadores e os valores angulares das articulações, o processo de dinâmica inversa pode ser realizado. Para tal feito, necessita-se além dos resultados cinemáticos, os valores de forças e momentos de reação no solo, possivelmente obtidos por meio de plataformas de força. Utilizando quaisquer interfaces do OpenSim, por meio da ferramenta de dinâmica inversa, torna-se possível o cálculo dos momentos articulares. Esse

procedimento é amplamente realizado no campo da biomecânica por se tratar de um método não invasivo para obtenção de variáveis que indicam a carga interna imposta nas articulações.

De forma resumida, Winter (2009) descreveu que a primeira etapa para obtenção dos momentos articulares são a medição do comprimento dos segmentos de cada participante. No OpenSim essas medidas são possíveis graças à distância entre os marcadores. Posteriormente, o centro de massa desses segmentos é calculado como proposto por Clauser, McConville et al. (1969). Contudo, no OpenSim, esses valores são baseados nas características de cada segmentos estabelecidos no modelo genérico e adaptado na etapa de escalonamento. Por fim, de acordo com o princípio da ação e reação, as forças do segmento distal são iguais em módulos, mas com sentidos opostos ao segmento proximal (Winter 2009). A partir disso, é calculado a força de reação na extremidade proximal dos segmentos e após a aplicação das equações de rotação, obtém-se os resultados dessa etapa: momentos articulares (Figura 7).

Figura 7. Fluxograma da ferramenta de Dinâmica Inversa no OpenSim. As caixas verdes representam os dados experimentais; o modelo escolhido é representado pela caixa laranja; o arquivo de configuração da ferramenta é indicado pela cor azul e as caixas roxas representam arquivos gerados por análises disponíveis OpenSim.

2.2.2.4 Otimização estática (*Static Optimization*)

Para evitar técnicas invasivas, abordagens como a otimização estática (em inglês, *static optimization* [SO]) e controle muscular computadorizado (em inglês, *computed muscle control* [CMC]), são oferecidas pelo OpenSim para obter as excitações musculares (controle dos

atuadores) ao longo de uma determinada tarefa. Assim, após este procedimento, obtêm-se os valores de força, controle e ativação muscular, no decorrer da tarefa analisada.

No presente documento, optou-se pela utilização da SO³, dado que essa técnica se apresenta mais eficiente para cálculo de concentrações como para a obtenção dos valores de força de reação articular. A SO usa como base a geometria muscular (individual para cada modelo musculoesquelético de escolha) e momento articular, permitindo que se estime os valores de força em um determinado instante (Brand, Crowninshield et al. 1982, Davy e Audu 1987, Edwards, Miller et al. 2016). Pode-se definir que a otimização estática é uma extensão do procedimento de dinâmica inversa, na qual os valores de forças aplicadas na articulação são divididos, quadro a quadro, nos músculos que compõem o modelo, mas, principalmente, respeitam as suas devidas proporções e critérios de performance (Hicks, Uchida et al. 2015).

Para o cálculo de otimização estática, são necessários os resultados das etapas descritas (cinemática e dinâmica inversa), escolha do modelo musculoesquelético, o qual contém os músculos disponíveis. Além disso, são utilizados arquivos com atuadores residuais (em inglês, *actuators*) que atuam como uma variável adicional de força quando não há produção de força mínima; ou seja, essa ferramenta é apenas utilizada quando necessário, visando garantir a ocorrência de uma aplicação mínima de força dos músculos (Figura 8).

É válido destacar que os cálculos da força de cada músculo são feitos a partir de uma fórmula de custo, impondo a equidade entre torques internos e externos. Entretanto, dado que essa técnica não utiliza dados eletromiográficos para o cálculo da força muscular, são utilizados *constrains* ou informações individuais para cada músculo como força máxima do músculo para melhorar a qualidade dos valores de força muscular. Este é um processamento simples e é uma técnica de baixo custo computacional, o que a torna atraente para mensurar a força muscular.

³ Entenda os procedimentos da SO:

<https://simtk-confluence.stanford.edu/display/OpenSim/Working+with+Static+Optimization>

Figura 8. Fluxograma da ferramenta de Otimização Estática do OpenSim. As caixas verdes representam os dados experimentais; o modelo escolhido é representado pela caixa laranja; o arquivo de configuração da ferramenta de otimização é indicado pela cor azul e as caixas roxas representam arquivos gerados por análises disponíveis OpenSim.

2.2.2.5 Força de reação articular (*Joint Reaction Force [JRF]*)

Após realizar todos os procedimentos, torna-se possível quantificar a força de reação articular. Embora essa quantificação possa ser feita utilizando métodos invasivos como próteses instrumentalizadas, a utilização de simulação computacional possibilita o cálculo dessas variáveis de modo não invasivo. Para isso, implementa-se a opção de análise de Reação Articular (em inglês, *Joint Reaction*⁴) no OpenSim.

Além dos valores de momentos articulares, essa ferramenta possibilita a quantificação da força resultante entre os ossos que compõem uma articulação. Desse modo, essa análise torna-se uma importante ferramenta para compreensão das cargas internas suportadas pelas estruturas que compõem a articulação, dado que o valor representado no ponto central da articulação representa o conjunto de forças aplicadas entre ossos, cartilagens e ligamentos.

Os valores de força articular podem ser expressos em diferentes formatos como força do segmento proximal aplicado no distal ou vice e versa. Além disso, o vetor de força pode ser referenciado tanto de acordo com o sistema de orientação do segmento distal, proximal ou solo. Como exemplo, no caso da articulação tibiofemoral abaixo (Figura 9), apresenta-se uma visão

⁴ Análise de Reação Articular:
<https://simtk-confluence.stanford.edu/display/OpenSim/Joint+Reactions+Analysis>

posterior da força de reação na articulação tibiofemoral aplicada e orientada no segmento distal (tíbia). Vale destacar que o padrão de referências escolhido pelos desenvolvedores do OpenSim os valores de reação articular são expressos no segmento distal e com referencial no solo.

Por fim, após a apresentação detalhada de todos os procedimentos utilizados, apresento abaixo um fluxograma contendo as etapas e procedimentos necessário para obtenção das variáveis de interesse (Figura 10):

Figura 9. As linhas representam a força de reação, normalizadas pela massa do participante, na articulação tibiofemoral. As linhas de cor preta e vermelha representam os valores de força de reação articular menores e maiores que maiores que 2 pesos corporais (sigla BW em inglês), respectivamente. Nesse caso, a força é aplicada e orientada no segmento distal (tíbia).

Figura 10. Fluxograma das etapas necessárias para obtenção das variáveis presentes neste documento

Referências

- Alentorn-Geli, E., G. D. Myer, H. J. Silvers, G. Samitier, D. Romero, C. Lázaro-Haro and R. Cugat (2009). "Prevention of non-contact anterior cruciate ligament injuries in soccer players. Part 1: Mechanisms of injury and underlying risk factors." Knee surgery, sports traumatology, arthroscopy **17**(7): 705-729.
- Allen, T. J. and U. Proske (2006). "Effect of muscle fatigue on the sense of limb position and movement." Experimental Brain Research **170**(1): 30-38.
- Arendt, E. and R. Dick (1995). "Knee injury patterns among men and women in collegiate basketball and soccer NCAA data and review of literature." The American journal of sports medicine **23**(6): 694-701.
- Arnold, A. S., S. S. Blemker and S. L. Delp (2001). "Evaluation of a deformable musculoskeletal model for estimating muscle-tendon lengths during crouch gait." Ann Biomed Eng **29**(3): 263-274.
- Bedo, B. L. S., D. S. Catelli, M. Lamontagne and P. R. P. Santiago (2020). "A custom musculoskeletal model for estimation of medial and lateral tibiofemoral contact forces during tasks with high knee and hip flexions." Comput Methods Biomech Biomed Engin (in press).
- Bedo, B. L. S., J. P. V. Manechini, M. Nunomura, R. P. Menezes and S. R. D. Silva (2019). "Injury Frequency in Handball Players: A Descriptive Study of Injury Pattern in São Paulo State Regional Teams." Motriz: Revista de Educação Física.
- Benjaminse, A., K. E. Webster, A. Kimp, M. Meijer and A. Gokeler (2019). "Revised Approach to the Role of Fatigue in Anterior Cruciate Ligament Injury Prevention: A Systematic Review with Meta-Analyses." Sports Med **49**(4): 565-586.
- Borotikar, B. S., R. Newcomer, R. Koppes and S. G. McLean (2008). "Combined effects of fatigue and decision making on female lower limb landing postures: central and peripheral contributions to ACL injury risk." Clin Biomech (Bristol, Avon) **23**(1): 81-92.
- Bourne, M. N., K. E. Webster and T. E. Hewett (2019). "Is Fatigue a Risk Factor for Anterior Cruciate Ligament Rupture?" Sports Med **49**(11): 1629-1635.
- Brand, R. A., R. D. Crowninshield, C. E. Wittstock, D. R. Pedersen, C. R. Clark and F. M. van Krieken (1982). "A model of lower extremity muscular anatomy." J Biomech Eng **104**(4): 304-310.
- Caine, D. J., C. G. Caine and K. J. Lindner (1996). "Epidemiology of Sports Injuries." The Nurse Practitioner **21**(9): 142.
- Cappozzo, A., F. Catani, U. D. Croce and A. Leardini (1995). "Position and orientation in space of bones during movement: anatomical frame definition and determination." Clin Biomech (Bristol, Avon) **10**(4): 171-178.
- Catelli, D. S., M. Wesseling, I. Jonkers and M. Lamontagne (2019). "A musculoskeletal model customized for squatting task." **22**(1): 21-24.
- Chappell, J. D., D. C. Herman, B. S. Knight, D. T. Kirkendall, W. E. Garrett and B. Yu (2005). "Effect of fatigue on knee kinetics and kinematics in stop-jump tasks." Am J Sports Med **33**(7): 1022-1029.
- Clauser, C. E., J. T. McConville and J. W. Young (1969). Weight, volume, and center of mass of segments of the human body, DTIC Document.
- Cochrane, J. L., D. G. Lloyd, A. Butfield, H. Seward and J. McGivern (2007). "Characteristics of anterior cruciate ligament injuries in Australian football." J Sci Med Sport **10**(2): 96-104.

- Cumps, E., E. Verhagen and R. Meeusen (2007). "Efficacy of a sports specific balance training programme on the incidence of ankle sprains in basketball." J Sports Sci Med **6**(2): 212-219.
- Damsgaard, M., J. Rasmussen, S. T. Christensen, E. Surma and M. de Zee (2006). "Analysis of musculoskeletal systems in the AnyBody Modeling System." Simulation Modelling Practice and Theory **14**(8): 1100-1111.
- Davy, D. T. and M. L. Audu (1987). "A dynamic optimization technique for predicting muscle forces in the swing phase of gait." Journal of Biomechanics **20**(2): 187-201.
- Delp, S. L., F. C. Anderson, A. S. Arnold, P. Loan, A. Habib, C. T. John, E. Guendelman and D. G. Thelen (2007). "OpenSim: open-source software to create and analyze dynamic simulations of movement." IEEE Trans Biomed Eng **54**(11): 1940-1950.
- Delp, S. L., J. P. Loan, M. G. Hoy, F. E. Zajac, E. L. Topp and J. M. Rosen (1990). "An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures." IEEE Trans Biomed Eng **37**(8): 757-767.
- Edwards, W. B., R. H. Miller and T. R. Derrick (2016). "Femoral strain during walking predicted with muscle forces from static and dynamic optimization." Journal of Biomechanics **49**(7): 1206-1213.
- Enoka, R. M., & Duchateau, J. (2016). Translating Fatigue to Human Performance. Med Sci Sports Exerc, 48(11), 2228-2238. doi: 10.1249/mss.0000000000000929
- Gandevia, S. C. (2001). "Spinal and supraspinal factors in human muscle fatigue." Physiol Rev **81**(4): 1725-1789.
- Griffin, L. Y., J. Agel, M. J. Albohm, E. A. Arendt, R. W. Dick, W. E. Garrett, J. G. Garrick, T. E. Hewett, L. Huston, M. L. Ireland, R. J. Johnson, W. B. Kibler, S. Lephart, J. L. Lewis, T. N. Lindenfeld, B. R. Mandelbaum, P. Marchak, C. C. Teitz and E. M. Wojtys (2000). "Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies." J Am Acad Orthop Surg **8**(3): 141-150.
- Griffin, L. Y., M. J. Albohm, E. A. Arendt, R. Bahr, B. D. Beynnon, M. Demaio, R. W. Dick, L. Engebretsen, W. E. Garrett, Jr., J. A. Hannafin, T. E. Hewett, L. J. Huston, M. L. Ireland, R. J. Johnson, S. Lephart, B. R. Mandelbaum, B. J. Mann, P. H. Marks, S. W. Marshall, G. Myklebust, F. R. Noyes, C. Powers, C. Shields, Jr., S. J. Shultz, H. Silvers, J. Slauterbeck, D. C. Taylor, C. C. Teitz, E. M. Wojtys and B. Yu (2006). "Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting, January 2005." Am J Sports Med **34**(9): 1512-1532.
- Hewett, T. E. (2000). "Neuromuscular and hormonal factors associated with knee injuries in female athletes." Sports medicine **29**(5): 313-327.
- Hewett, T. E., K. R. Ford and G. D. Myer (2005). "Anterior cruciate ligament injuries in female athletes - Part 2, a meta-analysis of neuromuscular interventions aimed at injury prevention." American Journal of Sports Medicine **34**(3): 490-498.
- Hewett, T. E., G. D. Myer and K. R. Ford (2001). "Prevention of anterior cruciate ligament injuries." Curr Womens Health Rep **1**(3): 218-224.
- Hewett, T. E., G. D. Myer, K. R. Ford, R. S. Heidt, Jr., A. J. Colosimo, S. G. McLean, A. J. van den Bogert, M. V. Paterno and P. Succop (2005). "Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study." Am J Sports Med **33**(4): 492-501.
- Hewett, T. E., M. V. Paterno and G. D. Myer (2002). "Strategies for enhancing proprioception and neuromuscular control of the knee." Clinical Orthopaedics and related research **402**: 76-94.
- Hicks, J. L., T. K. Uchida, A. Seth, A. Rajagopal and S. L. Delp (2015). "Is my model good enough? Best practices for verification and validation of musculoskeletal models and simulations of movement." J Biomech Eng **137**(2): 020905.

- Hiemstra, L. A., I. K. Lo and P. J. Fowler (2001). "Effect of fatigue on knee proprioception: implications for dynamic stabilization." J Orthop Sports Phys Ther **31**(10): 598-605.
- Hoff, G. L. and T. A. Martin (1986). "Outdoor and Indoor Soccer - Injuries among Youth Players." American Journal of Sports Medicine **14**(3): 231-233.
- Johnston, R. B., 3rd, M. E. Howard, P. W. Cawley and G. M. Losse (1998). "Effect of lower extremity muscular fatigue on motor control performance." Med Sci Sports Exerc **30**(12): 1703-1707.
- Koga, H., A. Nakamae, Y. Shima, J. Iwasa, G. Myklebust, L. Engebretsen, R. Bahr and T. Krosshaug (2010). "Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball." Am J Sports Med **38**(11): 2218-2225.
- Langevoort, G., G. Myklebust, J. Dvorak and A. Junge (2007). "Handball injuries during major international tournaments." Scandinavian Journal of Medicine & Science in Sports **17**(4): 400-407.
- Lattanzio, P. J., R. J. Petrella, J. R. Sproule and P. J. Fowler (1997). "Effects of fatigue on knee proprioception." Clin J Sport Med **7**(1): 22-27.
- Liederbach, M., I. J. Kremenik, K. F. Orishimo, E. Pappas and M. Hagins (2014). "Comparison of landing biomechanics between male and female dancers and athletes, part 2: Influence of fatigue and implications for anterior cruciate ligament injury." Am J Sports Med **42**(5): 1089-1095.
- Lindblad, B. E., K. Hoy, C. J. Terkelsen, H. E. Helleland and C. J. Terkelsen (1992). "Handball injuries. An epidemiologic and socioeconomic study." Am J Sports Med **20**(4): 441-444.
- Mair, S. D., A. V. Seaber, R. R. Glisson and W. E. Garrett, Jr. (1996). "The role of fatigue in susceptibility to acute muscle strain injury." Am J Sports Med **24**(2): 137-143.
- Mandelbaum, B. R., H. J. Silvers, D. S. Watanabe, J. F. Knarr, S. D. Thomas, L. Y. Griffin, D. T. Kirkendall and W. Garrett, Jr. (2005). "Effectiveness of a neuromuscular and proprioceptive training program in preventing anterior cruciate ligament injuries in female athletes: 2-year follow-up." Am J Sports Med **33**(7): 1003-1010.
- McLean, S. G., R. E. Fellin, N. Suedekum, G. Calabrese, A. Passerallo and S. Joy (2007). "Impact of fatigue on gender-based high-risk landing strategies." Med Sci Sports Exerc **39**(3): 502-514.
- Micheo, W., L. Hernandez and C. Seda (2010). "Evaluation, management, rehabilitation, and prevention of anterior cruciate ligament injury: current concepts." PM R **2**(10): 935-944.
- Moller, M., J. Attermann, G. Myklebust and N. Wedderkopp (2012). "Injury risk in Danish youth and senior elite handball using a new SMS text messages approach." Br J Sports Med **46**(7): 531-537.
- Noyes, F. R., L. A. Dunworth, T. P. Andriacchi, M. Andrews and T. E. Hewett (1996). "Knee hyperextension gait abnormalities in unstable knees - Recognition and preoperative gait retraining." American Journal of Sports Medicine **24**(1): 35-45.
- Olsen, O. E., G. Myklebust, L. Engebretsen and R. Bahr (2004). "Injury mechanisms for anterior cruciate ligament injuries in team handball a systematic video analysis." American Journal of Sports Medicine **32**(4): 1002-1012.
- Olsen, O. E., G. Myklebust, L. Engebretsen and R. Bahr (2006). "Injury pattern in youth team handball: a comparison of two prospective registration methods." Scandinavian Journal of Medicine & Science in Sports **16**(6): 426-432.
- Rajagopal, A., C. L. Dembia, M. S. DeMers, D. D. Delp, J. L. Hicks and S. L. Delp (2016). "Full-Body Musculoskeletal Model for Muscle-Driven Simulation of Human Gait." IEEE Trans Biomed Eng **63**(10): 2068-2079.

- Relph, N. (2015). The measurement of knee joint position sense, University of Salford.
- Sanches, F. G. and S. H. Borin (2009). "Lesões mais comuns no Handebol." Anuário da Produção Acadêmica Docente **2**(3): 233-240.
- Seth, A., J. L. Hicks, T. K. Uchida, A. Habib and C. L. Dembia (2018). "OpenSim: Simulating musculoskeletal dynamics and neuromuscular control to study human and animal movement." **14**(7): e1006223.
- Seth, A., M. A. Sherman, J. A. Reinbolt and S. L. Delp (2011). "OpenSim: a musculoskeletal modeling and simulation framework for in silico investigations and exchange." Procedia IUTAM **2**: 212-232.
- Skinner, H. B., M. P. Wyatt, M. L. Stone, J. A. Hodgdon and R. L. Barrack (1986). "Exercise-related knee joint laxity." Am J Sports Med **14**(1): 30-34.
- Stuart, M. J. and A. Smith (1995). "Injuries in Junior A ice hockey. A three-year prospective study." Am J Sports Med **23**(4): 458-461.
- Uchida, T. K. and S. L. Delp (2020). Biomechanics of Movement: The Science of Sports, Robotics, and Rehabilitation, MIT Press.
- van Mechelen, W., H. Hlobil and H. C. Kemper (1992). "Incidence, severity, aetiology and prevention of sports injuries. A review of concepts." Sports Med **14**(2): 82-99.
- Waldén, M., M. Hägglund, H. Magnusson and J. Ekstrand (2011). "Anterior cruciate ligament injury in elite football: a prospective three-cohort study." Knee Surg Sports Traumatol Arthrosc **19**(1): 11-19.
- Wedderkopp, N., M. Kaltoft, B. Lundgaard, M. Rosendahl and K. Froberg (1997). "Injuries in young female players in European team handball." Scand J Med Sci Sports **7**(6): 342-347.
- Wedderkopp, N., M. Kaltoft, B. Lundgaard, M. Rosendahl and K. Froberg (1999). "Prevention of injuries in young female players in European team handball. A prospective intervention study." Scand J Med Sci Sports **9**(1): 41-47.
- Whiting, W. C. and R. F. Zernicke (2009). Biomecânica funcional: e das lesões musculoesqueléticas, GUANABARA.
- Winter, D. A. (2009). Biomechanics and motor control of human movement, John Wiley & Sons.
- Wojtys, E. M., M. L. Beaulieu and J. A. Ashton-Miller (2016). "New perspectives on ACL injury: On the role of repetitive sub-maximal knee loading in causing ACL fatigue failure." J Orthop Res **34**(12): 2059-2068.
- Xia, R., X. Zhang, X. Wang, X. Sun and W. Fu (2017). "Effects of Two Fatigue Protocols on Impact Forces and Lower Extremity Kinematics during Drop Landings: Implications for Noncontact Anterior Cruciate Ligament Injury." Journal of healthcare engineering **2017**: 5690519-5690519.
- Zebis, M. K., L. L. Andersen, J. Bencke, M. Kjær and P. Aagaard (2009). "Identification of athletes at future risk of anterior cruciate ligament ruptures by neuromuscular screening." The American journal of sports medicine **37**(10): 1967-1973.

2.3 OBJETIVO GERAL

A presente tese objetivou-se em averiguar o efeito agudo da fadiga induzida por meio de um protocolo baseado em ações realizadas durante a prática de handebol, na biomecânica da articulação do joelho de atletas de handebol do sexo feminino.

2.4 ESTUDOS EMPÍRICOS

2.4.1 Estudo 1

Título: *The rapid recovery of vertical force propulsion production and postural sway after a specific fatigue protocol in female handball athletes.*

Objetivo: Investigar o efeito de um protocolo de fadiga baseado em ações frequentemente realizadas durante a prática do handebol no controle postural e produção de máxima força vertical. Além disso, foi quantificado o tempo necessário para a re-estabilização das variáveis de controle postural e força vertical.

2.4.2 Estudo 2

Título: *The effect of fatigue on hip and knee kinematics during single- and double-leg dynamic tasks: A study on female handball players.*

Objetivo: Investigar o efeito da fadiga induzida por meio de um protocolo específico para atletas de handebol na cinemática da articulação do joelho durante tarefas de aterrissagem bi- e uni-podal.

2.4.3 Estudo 3

Título: *A custom musculoskeletal model for estimation of medial and lateral tibiofemoral contact forces during tasks with high knee and hip flexions.*

Objetivo: Examinar o efeito da inclusão de um método de quantificação de forças mediais e laterais da articulação tibiofemoral em um modelo musculoesquelético que permita expressivos valores de flexão de quadril e joelho.

2.4.4 Estudo 4

Título: *BOPS: A Matlab toolbox to Batch Musculoskeletal Data Processing from OpenSim.*

Objetivo: Apresentar o *Batch OpenSim Processing Scripts* (BOPS), um *toolbox* elaborado em ambiente Matlab para otimizar o processamento das análises mais comuns realizadas no OpenSim (*Inverse Kinematics* - IK, *Inverse Dynamics* - ID, *Muscle Analysis* - MA, *Static Optimization* - SO, e *Joint Reaction Analysis* - JRA) possibilitando salvar os *outputs*, *logs*, *setups* e figuras, de modo automático.

2.4.5 Estudo 5

Título: *Effect of the fatigue in knee biomechanics during the sidestep cutting maneuver: a modeling approach.*

Objetivo: Investigar o efeito da fadiga, oriunda de um protocolo que simula ações de handebol, na biomecânica da articulação do joelho, utilizando uma abordagem de simulação e modelagem computacional.

2.5 DESAFIOS METODOLÓGICOS

Fundamentado nos métodos descritos nos tópicos anteriores, julgo pertinente destacar desafios relevantes para a elaboração da presente tese. Os desafios, embora indesejados, foram fundamentais para a melhoria na qualidade dos trabalhos apresentados, assim como para o crescimento acadêmico e profissional do candidato.

No que concerne ao crescimento acadêmico, foram tais desafios que possibilitaram o desenvolvimento de novos métodos e técnicas que foram fundamentais para possivelmente responder o questionamento central da tese.

Protocolo de fadiga

O primeiro desafio metodológico refere-se ao protocolo utilizado para indução à fadiga. Diversos métodos têm sido empregados para indução da mesma como esteiras ergométricas (Benjaminse, Habu et al. 2007, Zech, Steib et al. 2012, Steib, Zech et al. 2013), dinamômetro isocínético (Gribble, Hertel et al. 2004, Dieling, van der Esch et al. 2014), elevação unipodal (Zech, Steib et al. 2012), sequência de avanços unipodais (Gribble, Hertel et al. 2004), circuitos (Webster, Pietrosimone et al. 2016) e jogos reduzidos (Thorlund, Michalsik et al. 2008). Visando respeitar a especificidade do esporte, e possibilitar a realização de análises em ambiente laboratorial, elaborou-se um protocolo em forma de circuito a ser realizado dentro do laboratório. Deste modo, com foco nos estudos posteriores (Estudos [2](#) e [5](#)), esse protocolo possibilitou que as atletas fossem analisadas imediatamente após a detecção da fadiga.

Embora o protocolo de fadiga esteja descrito no [Estudo 1](#), convido os leitores a assistirem ao vídeo ilustrativo de uma participante realizando o protocolo de fadiga. O vídeo, que contém o exemplo de uma volta no protocolo na fase de aquecimento, pode ser acessado pelo link: “[Vídeo Protocolo de Fadiga](#)”⁵. Sua publicação foi autorizada pela voluntária.

⁵ Em caso de problemas em acessar o arquivo, segure a tecla “ctrl” e clique no link abaixo:

Quantificação da força durante o protocolo

A quantificação dos valores de força máxima de propulsão (F_{MAX}) ainda durante a realização do protocolo de fadiga descrito acima, era fundamental para a detecção rápida do estado de fadiga. Deste modo, visando não depender de softwares de aquisição e processamento de dados (ex: Nexus, Vicon, Oxford, UK), elaborou-se um aparato tecnológico que permitiu a quantificação das variáveis de força de forma “*ao vivo*” e com baixo custo computacional.

Para tal feito, foi utilizada uma plataforma de força Bertec (modelo #FP4060-NC-2000) conectados a um conversor Analógico-Digital (National Instruments #USB-6251-BNC, Austin, TX, USA) com entrada para oito canais analógicos e frequência de até 1,25 MS/s (Figura 1). A placa, conectada diretamente ao laptop, possibilitou a análise imediata dos dados brutos (F_x , F_y , F_z , M_x , M_y e M_z) por meio um código elaborado em ambiente Matlab ([Anexo 1](#)), tornando possível quantificar a força de propulsão máxima quase que instantaneamente. Desse modo, ainda durante a execução do protocolo, os pesquisadores conseguiram saber qual foi a porcentagem de redução de F_{MAX} ainda durante a execução do protocolo. Entretanto, para não enviesar o desempenho das atletas, esses valores não foram informados durante a execução do protocolo. Como sempre foi priorizado a capacidade da atleta de continuar; ou seja, no caso de a participante diminuir em ~5% a F_{MAX} (valor observado no [Estudo 1](#)), indicando a situação de fadiga, se ela ainda estivesse disposta para continuar, foi permitido. Ao final, embora quantificado os valores de força, a situação de fadiga sempre foi identificada pela incapacidade de a atleta continuar realizando o protocolo.

Figura 1. Conversor Analógico-Digital (National Instruments #USB-6251-BNC, Austin, TX, USA) utilizado para aquisição e conversão dos sinais de força e momento da plataforma de força.

Para quantificar a F_{MAX} , foi utilizado o salto contra movimento (em inglês, *countermovement jump*) (Figura 2). Para elucidação do salto, convido-os a acessar o link “[Vídeo Script Salto](#)”⁶ para observar o *script* em Matlab em funcionamento. Destaco que os códigos elaborados são os que calculam a força máxima de propulsão, considerada como o pico máximo de força antes de perder o contato com o solo, como a detecção automática do pico máximo de força (Figura 3). Esse método foi utilizado em todos os estudos que necessitaram quantificação da F_{MAX} no decorrer do protocolo de Fadiga (Estudos 1, 2 e 3).

⁶ Em caso de problemas em acessar o arquivo, segure a tecla “ctrl” e clique no link abaixo: <https://drive.google.com/file/d/1AVA7clyd2PgEhJ31YxF6awNK92v5PcMB/view?usp=sharing>
Se o problema persistir, por favor envie um e-mail para: bruno.bedo@usp.br ou PauloSantiago@usp.br

Figura 2. Ilustração do salto contra movimento (*countermovement jump*) realizado pelas atletas para quantificar a máxima força de propulsão.

Figura 3. Exemplo de um sinal de força durante a execução de um salto contra movimento (*countermovement jump*). O ponto vermelho representa a força máxima de propulsão antes do salto, enquanto o ponto azul representa a força pico de reação no solo.

Desenvolvimento de um modelo musculoesquelético e de uma ferramenta de otimização de análise de dados em OpenSim

Análises utilizando modelos musculoesqueléticos, devido às suas características não invasivas, são constantemente utilizadas para quantificação de cargas articulares (Delp, Anderson et al. 2007, Lai, Arnold et al. 2017, Mantovani e Lamontagne 2017, Mokhtarzadeh, Ewing et al. 2017, Catelli, Wesseling et al. 2019, Maniar, Bryant et al. 2020). Assim, como descrito nos tópicos anteriores, dentre as principais ferramentas que o OpenSim nos oferece, a análise de força de reação articular é uma das principais ferramentas para compreensão dos valores de força internos em uma articulação.

No que concerne a articulação tibiofemoral, os modelos musculoesqueléticos normalmente quantificam a força em um único ponto na articulação (Rajagopal, Dembia et al. 2016, Catelli, Wesseling et al. 2019) (Figura 4). Contudo, existem outros modelos que possibilitam quantificar a força de contato dos côndilos laterais e mediais, fornecendo dois vetores de força na mesma articulação (Lerner, DeMers et al. 2015). Entretanto, embora esses modelos sejam mais robustos do ponto de vista de análises articulares, a amplitude de movimento para as articulações do quadril e joelho são limitados; ou seja, esses modelos são limitados no ponto de vista de utilizá-los em atividades que requeiram expressivos valores de flexão de joelho e quadril como saltos, aterrissagens, agachamentos e mudanças de direção.

Assim, dado que um dos objetivos da presente tese foi quantificar o efeito da fadiga na carga interna da articulação, mais especificamente nos valores de força de reação articular nos côndilos lateral e medial da articulação tibiofemoral realização da mudança de direção ([Estudo 2](#)), encontramos-nos na necessidade de desenvolver um modelo que possibilitasse tais análises: quantificação dos componentes mediais e laterais de força de reação articular durante tarefas com expressivos valores de flexão de joelho e quadril. Deste modo, um dos grandes desafios da presente tese foi a elaboração de um modelo que tornasse possível tais análises.

O leitor encontrará informações detalhadas do processo de elaboração do modelo no [Capítulo 4](#); entretanto, com intuito de elucidar a etapa de superação desse desafio, na figura abaixo é possível observar um exemplo dos resultados do modelo desenvolvido para quantificar as forças mediais e laterais da articulação tibiofemoral durante um agachamento.

Figura 4. O gráfico da esquerda representa a força vertical de reação articular de um modelo que quantifica a força em um único ponto (preto) e a soma das componentes laterais/mediais (vermelho tracejado). Os gráficos da direita representam a força vertical no componente medial (topo) e lateral (abaixo) durante um agachamento para o modelo desenvolvido.

Embora o OpenSim seja uma excelente ferramenta para análises biomecânicas, são necessárias um número expressivo de etapas que, casualmente, podem resultar em erros. O software possibilita apenas o processamento de uma única tentativa que pode carecer de até duas horas de análises. Assim, ferramentas que otimizem o processamento podem auxiliar no processamento de dados, evitando erros humanos ou otimizando o processamento de dados.

Mantoan e Reggiani (2015) desenvolveram uma versão beta de um *toolbox*, em linguagem Matlab, intitulado *Batch OpenSim Processing Scripts* (BOPS). Essa ferramenta é um conjunto de *scripts* que possibilitam quantificar ângulos e momentos articulares, como

forças musculares de maneira idêntica ao OpenSim, dado que a ferramenta utiliza a *Application Programming Interface* (API) do OpenSim⁷. Entretanto, a ferramenta foi lançada em uma versão beta (denominada BOPsv0.9), na qual não foi possível calcular as forças de reação articular e possuía erros (*bugs*) que impossibilitavam a utilização massiva da ferramenta. Portanto, vimos uma possibilidade de auxiliar o meio acadêmico atualizando a ferramenta e, principalmente, otimizando o processo de análise.

Após realizar diversas atualizações na ferramenta, corrigindo *bugs* e inserindo a possibilidade de calcular a força de reação articular, contatamos as professoras Dra. Alice Mantoan e Dra. Monica Reggiane para que divulgássemos a nova versão da ferramenta, intitulado BOPsv2.0 (informações adicionais serão apresentadas no [Capítulo 6](#)). Por fim, a ferramenta encontra-se disponível para uso gratuito e tem auxiliado pesquisadores de diferentes laboratórios, dado que ela não requisita níveis avançados de linguagens de programação e possibilita análises idênticas às realizadas diretamente no OpenSim. Destaco que a ferramenta é licenciada em *Apache Licence, Version 2.0 (the “Licence”)*. Convido os leitores para assistirem um vídeo tutorial da utilização do BOPS: “[Vídeo Tutorial BOPS](#)”⁸.

Ressalto que para realização de todos os estudos empíricos apresentados na presente tese, foram necessárias adaptações que resultaram na possibilidade de realizar uma análise mais robusta e assertiva. Alguns estudos desenvolvidos permitem que pesquisadores de outras instituições utilizem ferramentas que poderão auxiliar e facilitar futuras análises.

Statistical parametric mapping

Ao decorrer dos estudos que compõem a presente tese, diferentes tarefas foram

⁷ OpenSim API para Matlab:

<https://simtk-confluence.stanford.edu/display/OpenSim/Scripting+with+Matlab>

⁸ Em caso de problemas em acessar o arquivo, segure a tecla “ctrl” e clique no link abaixo:

<https://drive.google.com/file/d/15QDnAHFbf-UFfPgZQil3rC2QDdP1DKdK/view?usp=sharing> ou

https://simtk.org/docman/?group_id=1038

Se o problema persistir, por favor envie um e-mail para: bruno.bedo@usp.br ou paulosantiago@usp.br

realizadas para avaliar a biomecânica dos membros inferiores. Deste modo, visando compreender os efeitos da fadiga durante os períodos de análises, buscaram-se métodos estatísticos que possibilitassem comparar uma variável em condições diferentes de modo contínuo, permitindo comparações quadro a quadro ao decorrer da janela de análise. Para isso optou-se na utilização de métodos de análises contínuas.

A utilização desses métodos em variáveis biomecânicas mostra-se uma importante ferramenta para compreensão não apenas de possíveis diferenças mas, além disso, dos instantes em que essas diferenças acontecem. Isso deve-se ao fato que a maioria das análises contínuas possibilitam analisar todo o conjunto de dados (desde que o conjunto de dados possua o mesmo tamanho), fato que não acontece em análises discretas, quando é utilizado, por exemplo, valores máximos, médios ou mínimos de uma determinada métrica.

Entre as análises contínuas, a Análise Paramétrica Mapeada (em inglês, *Statistical Parametric Mapping* [SPM]), encontra-se entre um dos principais métodos para análises contínuas em variáveis biomecânicas. Embora a sua aplicação do campo da biomecânica seja relativamente nova, principalmente no cenário nacional, esse método foi desenvolvido na década de 1990 para análises de neuroimagem para gerar mapas estatísticos paramétricos que pudessem determinar diferenças relevantes e efeitos regionais específicos nas amostras de interesse (Friston, Holmes et al. 1995, Friston, Ashburner et al. 2011, Ashburner 2012)

A primeira aparição da utilização deste método no campo da biomecânica foi no estudo desenvolvido por Pataky, Caravaggi et al. (2008) , na qual os pesquisadores apresentaram essa nova técnica em análises de pressão plantar com a velocidade da marcha. Pataky (2010) e Pataky, Robinson et al. (2013) apresentaram a possibilidade de utilizar esse método em diversas variáveis biomecânicas independentemente do número de dimensões e domínios espaço temporais. Como exemplo, considerando d o número de dimensões e n o número de domínios, abaixo apresento algumas das possibilidades de análises (Pataky 2010):

- Valores angulares de uma determinada articulação: $d = 1$; $n = 1$
- Componentes da trajetória de força (xyz): $d = 3$; $n = 1$
- Pressão plantar: $d = 1$; $n = 2$
- Campos de tensão em estruturas ósseas: $d = 6$; $n = 3$

Embora essa ferramenta pareça complexa, repositórios públicos são disponíveis⁹ para possibilitar a realização das análises sem a necessidade de compreensão profunda de estatística. Na documentação disponível é possível encontrar uma vasta possibilidade de análises possíveis, assim como exemplos de como os códigos devem ser elaborados. Até o presente momento, os códigos estão disponíveis nas linguagens de Matlab¹⁰ e Python¹¹.

Na presente tese, as variáveis investigadas serão abordadas de modo unidimensional e no domínio do tempo. Assim, visando elucidar os leitores dos métodos empregados nos estudos [2](#) e [5](#), apresento uma figura para explicar a visualização e interpretação de gráfico de SPM na comparação entre duas condições usando um teste-t para amostras dependentes.

⁹ SPM site: <https://spm1d.org>

¹⁰ Repositório SPM para Matlab: <https://github.com/Otodd0000/spm1dmatlab>

¹¹ Repositório SPM para Python: <https://github.com/Otodd0000/spm1d/>

Figura 5. Valores de abdução e adução da articulação do joelho durante a fase de contato na mudança de direção (cima). As linhas pretas e vermelhas representam os valores de média e desvio padrão das condições 1 e 2, respectivamente. O ciclo investigado (ilustração dos membros inferiores) e os resultados da análise pareada realizada por meio do teste-t ($SPM\{t\}$) são apresentados na figura debaixo. As linhas tracejadas e números vermelhos representam os limites críticos para ocorrência de diferença significativa ($t^*=3.33$). Por fim, a área sombreada representa o período em que houve diferença significativa que, no caso, foi entre 0 e 10% do ciclo investigado com um valor de significância de $p=0.03$.

Referências

- [1] S.C. Gandevia, Spinal and supraspinal factors in human muscle fatigue, *Physiological reviews* 81(4) (2001) 1725-89.
- [2] J.D. Chappell, D.C. Herman, B.S. Knight, D.T. Kirkendall, W.E. Garrett, B. Yu, Effect of fatigue on knee kinetics and kinematics in stop-jump tasks, *Am J Sports Med* 33(7) (2005) 1022-9.
- [3] L.A. Hiemstra, I.K. Lo, P.J. Fowler, Effect of fatigue on knee proprioception: implications for dynamic stabilization, *The Journal of orthopaedic and sports physical therapy* 31(10) (2001) 598-605.
- [4] S. Steib, A. Zech, C. Hentschke, K. Pfeifer, Fatigue-induced alterations of static and dynamic postural control in athletes with a history of ankle sprain, *J Athl Train* 48(2) (2013) 203-8.
- [5] S. Dieling, M. van der Esch, T.W. Janssen, Knee joint proprioception in ballet dancers and non-dancers, *Journal of dance medicine & science : official publication of the International Association for Dance Medicine & Science* 18(4) (2014) 143-8.
- [6] F.A. Barbieri, S.T. Rodrigues, P.F. Polastri, R.A. Barbieri, P.H.A. de Paula, F. Milioni, et al., High intensity repeated sprints impair postural control, but with no effects on free throwing accuracy, in under-19 basketball players, *Human movement science* 54 (2017) 191-196.
- [7] M. Ahmadi, A. Nodehi Moghadam, M. Bakhshi, T. Rezaeian, G. Pahnabi, The duration effects of lumbar extensor muscle fatigue on postural control in soccer players, *Medical journal of the Islamic Republic of Iran* 31 (2017) 112.
- [8] P.A. Gribble, J. Hertel, Effect of lower-extremity muscle fatigue on postural control, *Archives of physical medicine and rehabilitation* 85(4) (2004) 589-92.
- [9] A. Zech, S. Steib, C. Hentschke, H. Eckhardt, K. Pfeifer, Effects of localized and general fatigue on static and dynamic postural control in male team handball athletes, *Journal of strength and conditioning research* 26(4) (2012) 1162-8.
- [10] Z.G. Fox, J.P. Mihalik, J.T. Blackburn, C.L. Battaglini, K.M. Guskiewicz, Return of postural control to baseline after anaerobic and aerobic exercise protocols, *J Athl Train* 43(5) (2008) 456-63.
- [11] L. Guidetti, E. Franciosi, M.C. Gallotta, G.P. Emerenziani, C. Baldari, Postural control after a prolonged treadmill run at individual ventilatory and anaerobic threshold, *Journal of sports science & medicine* 10(3) (2011) 515-9.
- [12] D. Dickin, J. Doan, Postural stability in altered and unaltered sensory environments following fatiguing exercise of lower extremity joints, *Scandinavian journal of medicine & science in sports* 18(6) (2008) 765-772.
- [13] P.A. Gribble, J. Hertel, C.R. Denegar, W.E. Buckley, The Effects of Fatigue and Chronic Ankle Instability on Dynamic Postural Control, *J Athl Train* 39(4) (2004) 321-329.
- [14] K.A. Webster, B.G. Pietrosimone, P.A. Gribble, Muscle Activation During Landing Before and After Fatigue in Individuals With or Without Chronic Ankle Instability, *J Athl Train* (2016).
- [15] J.B. Thorlund, L.B. Michalsik, K. Madsen, P. Aagaard, Acute fatigue-induced changes in muscle mechanical properties and neuromuscular activity in elite handball players following a handball match, *Scand J Med Sci Sports* 18(4) (2008) 462-72.
- [16] T.E. Hewett, K.R. Ford, G.D. Myer, Anterior cruciate ligament injuries in female athletes - Part 2, a meta-analysis of neuromuscular interventions aimed at injury prevention, *Am J Sport Med* 34(3) (2005) 490-498. <Go to ISI>://WOS:000235449900020.
- [17] G. Langevoort, G. Myklebust, J. Dvorak, A. Junge, Handball injuries during major international tournaments, *Scandinavian journal of medicine & science in sports* 17(4) (2007) 400-407. <Go to ISI>://WOS:000248481300015.

- [18] M.J. Stuart, A. Smith, Injuries in Junior A ice hockey. A three-year prospective study, *Am J Sports Med* 23(4) (1995) 458-61.
- [19] C. Foster, J.A. Florhaug, J. Franklin, L. Gottschall, L.A. Hrovatin, S. Parker, et al., A new approach to monitoring exercise training, *Journal of strength and conditioning research* 15(1) (2001) 109-15.
- [20] M. Duarte, S.M.S.F. Freitas, Revisão sobre posturografia baseada em plataforma de força para avaliação do equilíbrio, *Brazilian Journal of Physical Therapy* 14 (2010) 183-192. http://www.scielo.br/scielo.php?script=sci_arttext&pid=S1413-35552010000300003&nrm=iso.
- [21] R. Moraes, B.L.S. Bedo, L.O. Santos, R.A. Batistela, P.R.P. Santiago, E. Mauerberg-deCastro, Additional Haptic Information Provided by Anchors Reduces Postural Sway in Young Adults Less Than Does Light Touch, *Frontiers in neuroscience* 12 (2018) 346.
- [22] M. Nocella, B. Colombini, G. Benelli, G. Cecchi, M.A. Bagni, J. Bruton, Force decline during fatigue is due to both a decrease in the force per individual cross-bridge and the number of cross-bridges, *The Journal of physiology* 589(Pt 13) (2011) 3371-3381. <https://www.ncbi.nlm.nih.gov/pubmed/21540343>
<https://www.ncbi.nlm.nih.gov/pmc/PMC3145945/>.
- [23] R.H. Fitts, Cellular mechanisms of muscle fatigue, *Physiological reviews* 74(1) (1994) 49-94.
- [24] F.O. Pires, Thomas Kuhn's 'Structure of Scientific Revolutions' applied to exercise science paradigm shifts: example including the Central Governor Model, *Br J Sports Med* 47(11) (2013) 721-2.
- [25] T.D. Noakes, Physiological models to understand exercise fatigue and the adaptations that predict or enhance athletic performance, *Scandinavian journal of medicine & science in sports* 10(3) (2000) 123-45.
- [26] S. Boyas, C. McGown, Y. Lajoie, M. Bilodeau, Changes in posture associated with postural control impairment induced by ankle plantarflexor fatigue during unipedal quiet standing, *Neuroscience letters* 547 (2013) 48-52.
- [27] T. Paillard, Effects of general and local fatigue on postural control: a review, *Neuroscience and biobehavioral reviews* 36(1) (2012) 162-76.
- [28] A.B.R. Pereira, R. Moraes, The Use of the Anchor System Reduces Postural Sway During Upright Standing Irrespective of Plantar Flexors Muscle Fatigue in Young and Older Adults, *Motor control* 22(3) (2018) 338-357.
- [29] K. Koyama, J. Yamauchi, Altered postural sway following fatiguing foot muscle exercises, *PLoS One* 12(12) (2017) e0189184.
- [30] T.J. Allen, U. Proske, Effect of muscle fatigue on the sense of limb position and movement, *Experimental Brain Research* 170(1) (2006) 30-38. <https://doi.org/10.1007/s00221-005-0174-z>.
- [31] N. Forestier, N. Teasdale, V. Nougier, Alteration of the position sense at the ankle induced by muscular fatigue in humans, *Med Sci Sports Exerc* 34(1) (2002) 117-22.
- [32] R. Lepers, A.X. Bigard, J.-P. Diard, J.-F. Gouteyron, C.Y. Guezennec, Posture control after prolonged exercise, *European journal of applied physiology and occupational physiology* 76(1) (1997) 55-61. <http://dx.doi.org/10.1007/s004210050212>.
- [33] S.D. Bot, A.P. Hollander, The relationship between heart rate and oxygen uptake during non-steady state exercise, *Ergonomics* 43(10) (2000) 1578-92.
- [34] L. Janssens, S. Brumagne, K. Polspoel, T. Troosters, A. McConnell, The effect of inspiratory muscles fatigue on postural control in people with and without recurrent low back pain, *Spine* 35(10) (2010) 1088-94.

- [35] J.W. Feuerbach, M.D. Grabiner, Effect of the aircast on unilateral postural control: amplitude and frequency variables, *The Journal of orthopaedic and sports physical therapy* 17(3) (1993) 149-54.
- [36] M.L. Beaulieu, M. Lamontagne, L. Xu, Gender differences in time-frequency EMG analysis of unanticipated cutting maneuvers, *Med Sci Sports Exerc* 40(10) (2008) 1795-804.
- [37] A. Nardone, J. Tarantola, A. Giordano, M. Schieppati, Fatigue effects on body balance, *Electroencephalography and clinical neurophysiology* 105(4) (1997) 309-20.
- [38] B.K. Springer, D.M. Pincivero, The effects of localized muscle and whole-body fatigue on single-leg balance between healthy men and women, *Gait & posture* 30(1) (2009) 50-4.
- [39] V. Agostini, E. Chiaramello, L. Canavese, C. Bredariol, M. Knaflitz, Postural sway in volleyball players, *Human movement science* 32(3) (2013) 445-56.
- [40] F. Baghbani, L.J. Woodhouse, A.A. Gaeini, Dynamic Postural Control in Female Athletes and Nonathletes After a Whole-Body Fatigue Protocol, *Journal of strength and conditioning research* 30(7) (2016) 1942-7.
- [41] V. Sakellari, A.M. Bronstein, Hyperventilation effect on postural sway, *Archives of physical medicine and rehabilitation* 78(7) (1997) 730-6.
- [42] T. Tanaka, N. Hashimoto, M. Nakata, T. Ito, S. Ino, T. Ifukube, Analysis of toe pressures under the foot while dynamic standing on one foot in healthy subjects, *The Journal of orthopaedic and sports physical therapy* 23(3) (1996) 188-93.
- [43] M.G. Carpenter, J.S. Frank, D.A. Winter, G.W. Peysar, Sampling duration effects on centre of pressure summary measures, *Gait & posture* 13(1) (2001) 35-40.

CAPÍTULO 2

ESTUDO 1

The rapid recovery of vertical force propulsion production and postural sway after a specific fatigue protocol in female handball athletes

Bruno Luiz Souza Bedo; Dayanne Rodrigues Pereira; Renato Moraes; Carlos Augusto Kalva-Filho; Tenyson Will-de-Lemos; Paulo Roberto Pereira Santiago

The rapid recovery of vertical force propulsion production and postural sway after a specific fatigue protocol in female handball athletes

Bruno Luiz Souza Bedo ^{A B}; Dayanne Rodrigues Pereira ^{A C}; Renato Moraes ^{A B}; Carlos Augusto Kalva-Filho ^B; Tenysson Will-de-Lemos ^A; Paulo Roberto Pereira Santiago ^{A B}

^ABiomechanics and Motor Control Laboratory, School of Physical Education and Sport of Ribeirão Preto, University of São Paulo.

^BRehabilitation and Functional Performance Graduate Program, Ribeirão Preto Medical School, University of São Paulo.

^CRibeirão Preto Medical School, University of São Paulo, Ribeirão Preto, São Paulo, Brazil.

A versão publicada pode ser acessada clicando [aqui](#) ou pelo QR code abaixo.¹²

Favor citar:

Bedo, B. L. S., Pereira, D. R., Moraes, R., Kalva-Filho, C. A., Will-de-Lemos, T., & Santiago, P. R. P. (2020). The rapid recovery of vertical force propulsion production and postural sway after a specific fatigue protocol in female handball athletes. *Gait & Posture*, 77, 52-58.

¹² Em caso de dificuldade para acessar o arquivo, segure a tecla “ctrl” e clique no link abaixo.
Link: <https://www.sciencedirect.com/science/article/abs/pii/S0966636220300436?via%3Dihub>
Se o problema persistir, por favor envie um e-mail para: bruno.bedo@usp.br ou paulosantiag@usp.br.

Abstract

Background: Muscle fatigue is characterized by a decrease in muscle performance, accompanied by an increase in perceived exertion to produce a desired amount of force. The duration of the fatigue effects is not completely clear regarding postural sway and force production in athletes.

Research question: What is the effect of a specific fatigue protocol in postural sway and force production in female handball athletes and the recovery time necessary for these variables?

Methods: Twenty female handball athletes participated in this study. Specific handball actions composed the fatigue protocol in the format of a circuit with the gradual increment of laps. They stood upright in one-leg posture on a force plate. For the maximum propulsion force (F_{MAX}), they performed a countermovement jump on the force plate. The center of pressure (COP) and the F_{MAX} were obtained at baseline, immediately after the exhaustion due to the fatigue protocol and every minute during the first 10 minutes of the recovery phase.

Results: The F_{MAX} decreased ~9.5% after the fatigue protocol, whereas the COP area increased 224%. During the recovery phase, the F_{MAX} and the COP area returned to baseline values after five minutes.

Significance: The first five minutes immediately after the fatigue protocol represent the time interval where the fatigue effects are still present. Future studies willing to investigate the effects of fatigue in athletes should perform the analysis within this short time-window.

Keywords: fatigue; postural sway; force; handball.

1. Introduction

Muscle fatigue results from a complex interaction between different body systems and is defined as an exercise-induced reduction in the ability to exert muscle force or power (Gandevia, 2001). Among several deleterious effects for movement performance, fatigue relates to modification on body homeostasis, especially in the proprioceptive system (Chappell et al., 2005; Hiemstra et al., 2001), which in turn contributes to decreasing joint stability and compromises the ligaments' stability (Hiemstra et al., 2001; Steib et al., 2013). In addition, postural control is impaired (Dieling et al., 2014; Steib et al., 2013). These deleterious effects have been observed in athletes of different modalities (Ahmadi, Nodehi Moghadam, Bakhshi, Rezaeian, & Pahnabi, 2017; Barbieri et al., 2017; Gribble & Hertel, 2004; Zech et al., 2012).

However, the effects of fatigue on balancing tasks are not entirely understood (Zech et al., 2012), as well as the time of recovery of the postural sway and force parameters. Fox, Mihalik, Blackburn, Battaglini, and Guskiewicz (2008) observed the adverse effects of fatigue on postural stability until the thirteenth minute using either aerobic or anaerobic exercises. Differently, Guidetti, Franciosi, Gallotta, Emerenziani, and Baldari (2011) found that the impact of fatigue remained until the tenth minute after exercising at the anaerobic threshold and until the fifth minute after exercising at the post-ventilatory threshold. Finally, Dickin and Doan (2008) observed that the recovery of the postural control system remained up to 30 minutes after the fatigue protocol. Although the influence of fatigue on the postural control is quite clear, results are divergent about the magnitude of the adverse effects and the time-course of the recovery phase. The protocol used to induce fatigue must consider the specificity of the sport modality, as the time-course of the recovery phase is specific to the fatigue protocol used.

Different fatigue protocols have been based on treadmill (Steib et al., 2013; Zech et al., 2012), isokinetic dynamometer (Dieling et al., 2014; Gribble et al., 2004), one-leg elevation (Zech et al., 2012), landing (Gribble et al., 2004), circuits (Webster et al., 2016), and small

games (Thorlund et al., 2008) tasks. Movements presented in a handball game have not been used in previous fatigue protocols. This lack of specificity prevents the understanding of fatigue effects in court sports. Thus, considering the high incidence of lower limb injuries in handball athletes (Hewett et al., 2005; Langevoort et al., 2007), possibly related to fatigue (Chappell et al., 2005; Stuart & Smith, 1995), it becomes important to investigate the effect of fatigue in this population. Besides, it is necessary to know the duration of the-fatigue effects on postural control such that in future studies will be possible to investigate their effects in joint instability.

Therefore, the purpose of this study was to investigate the effect of a fatigue protocol in the postural control and maximum force propulsion and quantify the recovery time necessary for these variables in female handball athletes. The fatigue protocol was elaborated to be performed in the laboratory setup and composed by movements present in handball practice. We hypothesized that fatigue would reduce the capacity to produce force and increase the postural sway; however, a few minutes would be necessary to return these parameters to baseline.

2. Methods

2.1 Participants

Twenty female handball athletes (21.9 ± 3.4 years; 63.5 ± 9.1 kg; 1.76 ± 0.07 m; 7.2 ± 3.2 years of experience; 240 ± 180.4 minutes of practice/week) without lower limb injuries in the last six months participated in this study. This sample size allowed a statistical power of 0.97 ($\alpha=0.05$; effect size=0.83) between baseline values and those observed in the fatigue state. This analysis was made using G*Power (version 3.1.9.2, Dusseldorf, Germany) and considering the maximum propulsion force as the primary output based on a previous study (Thorlund et al., 2008). The local Ethics Committee approved all the experimental procedures.

2.2 Fatigue Protocol

The protocol was elaborated to be performed in the laboratory setup. It focused on movements performed during handball practice, such as sprint, sidestep cutting, jumping, landing, lateral and back shifting, among others (see Figure 1 caption for details). The protocol was divided into rounds with a gradual increase in volume, and the athletes should perform each round as fast as possible. The round number was the number of laps in the circuit (i.e., round I [1 lap]; round II [2 laps]; and so on). At the end of each lap, the heart rate (HR, Polar V800TM), the time of effort, and the rate of perceived exertion (RPE using the Foster scale [0-10] (Foster et al., 2001)) were assessed. At the end of each round, the propulsion force was quantified by a countermovement jump performed on a force plate. Participants were instructed to jump as high as possible without flexing the knees and hips during the flight phase. The participants completed two laps in the circuit at a lower intensity for acclimation and warm-up. The state of fatigue was characterized by the inability of the participant to continue performing the protocol.

Figure 1. Illustration of the fatigue protocol based on specific movements performed by handball athletes. The athletes should perform the following actions in all of the stations as fast as possible: (A) Acceleration and sidestep cutting maneuver for the right and run until the end of the 3-m distance; receive a pass (at the end of the 3-m length) from a partner positioned 2.5 meters away (Partner 1); backshift until the 90°; run 3 m for the left side, and perform an accurate throw to a partner positioned 2.5 meters away (Partner 2); this task was performed twice per lap; (B) Ten double-leg lateral jumping/landing over a barrier (20-cm high), performed as fast as possible; (C) Sprint; (D) Double-leg plyometric jumps, performed as fast as possible; (E) Lateral shifting; (F) Sprint; (G) Back Shifting; and (H) Sprint with a 90° change of direction. The protocol was divided into rounds, where the round number represents the number of laps in the circuit (i.e., round I [1 lap]; round II [2 laps]; round III [3 laps]; and so on) until the participant is unable to continue performing the protocol.

2.3 Fatigue Protocol

The assessment of postural sway and force production was performed at baseline, immediately after the exhaustion due to the fatigue protocol, and at every minute during the first 10 minutes of the recovery phase. The participants stood upright on a force plate (40 × 60 cm, Bertec, Columbus, USA) in a one-leg posture using the non-dominant limb as the support limb, determined by asking the leg used to take off in the handball shoot (Zech et al., 2012). They were instructed to stand as quietly as possible for 30 s with eyes opened, looking to a target positioned 3 m away. On the same force plate, the participants executed a countermovement jump to compute the maximum propulsion force (F_{MAX}). The force plate data were collected using an A/D converter (National Instruments #USB-6251-BNC, Austin, TX, USA), sampled at 5000 Hz using the software Matlab® (Mathworks INC. Natick, MA, USA).

2.4. Data analyses

The raw force and moment data were filtered with a low-pass fourth-order Butterworth

filter with a cut-off frequency of 5 Hz. The center of pressure (COP) displacement was calculated using the force and moment data. The following variables were computed using the COP displacement (Duarte & Freitas, 2010): ellipse area containing 95% of the COP data points (Area), mean sway speed (MSS), root mean square (RMS) and the frequency at 80% of the power spectrum density (F80) (Moraes et al., 2018). Except for Area, all remaining variables were computed in AP and ML directions.

The baseline data was computed as the mean of three countermovement jumps, separated by an interval of 60 s between each jump, to increase the robustness of the F_{MAX} . After the fatigue protocol, the participants performed only one jump per minute during the recovery phase. Force data were filtered with a low-pass fourth-order Butterworth filter with a cut-off of 10 Hz. The F_{MAX} was determined as the first peak of the vertical component of the ground reaction force during the countermovement jump execution. The F_{MAX} was normalized by the participants' total body weight, and the percentage of F_{MAX} alteration was quantified between the baseline and the measures after the fatigue protocol.

2.5. Statistical analyses

The first one-way analysis of variance (ANOVA) for repeated measures (baseline) assessed the fatigue parameters (Time, F_{MAX} , HR, and RPE) between the baseline, Round 1, and immediately after the fatigue protocol. The effect of fatigue on the dependent variables during the recovery phase was investigated using a second ANOVA for repeated measures. When the ANOVA identified main effects, the Least Significant Difference (LSD) *post-hoc* analysis was carried out to determine the pairwise differences. The significance level was set at $\alpha < 0.05$.

3. Results

Each participant executed a particular number of laps to achieve fatigue, ranging between three and ten (4.6 ± 2.1 laps or 203 ± 84.9 s). Significant increases were observed for HR ($F_{3,57}=392.72$; $p < 0.001$), RPE ($F_{3,57}=366.11$; $p < 0.001$), and the time to complete the last lap ($F_{2,38}=37.35$; $p < 0.0001$) (Table 1). After the fatigue, during the recovery phase, the ANOVA showed a progressive decrease in RPE ($F_{11,209}=90.16$; $p < 0.0001$) and HR ($F_{11,209}=166.14$; $p < 0.0001$), but they never returned to baseline values (Figure 2).

Table 1. Mean and standard deviation ($\pm$) of the fatigue parameters.

	Baseline	Round 1	Fatigue State
Time (s)	-	39.65 ± 1.49	56.85 ± 1.68^a
F_{MAX} (%)	100	93.47 ± 6.90^b	90.53 ± 7.82^{ab}
Hear Rate (bpm)	80.15 ± 5.71	161.85 ± 19.65^b	192.90 ± 4.45^{ab}
RPE	0 ± 0	2.65 ± 0.21^b	10 ± 0^{ab}

Note: Rated Perceived Exertion (RPE); ^a Different from Round 1; ^b Different from Rest; $p < 0.05$.

The F_{MAX} reduced immediately after the fatigue protocol compared to baseline ($F_{3,57}=7.75$; $p=0.01$). Immediately after the fatigue protocol (0 min), the F_{MAX} equaled to 90.5% ($p=0.001$), and this reduction lasted until the forth minute: 1 min (94.1%, $p < 0.0001$), 2 min (94.0%, $p < 0.0001$), 3 min (94.8%, $p=0.004$), and 4 min (94.6%, $p=0.023$). It returned to baseline values at the fifth minute (97.3%; $p=0.137$), but during the 8th and 9th minutes, the F_{MAX} decreased again (Figure 2).

The F_{MAX} reduced immediately after the fatigue protocol compared to baseline ($F_{3,57}=7.75$; $p=0.01$). Immediately after the fatigue protocol (0 min), the F_{MAX} equaled to 90.5% ($p=0.001$), and this reduction lasted until the forth minute: 1 min (94.1%, $p < 0.0001$), 2 min (94.0%, $p < 0.0001$), 3 min (94.8%, $p=0.004$), and 4 min (94.6%, $p=0.023$). It returned to baseline values at the fifth minute (97.3%; $p=0.137$), but during the 8th and 9th minutes, the

F_{MAX} decreased again (Figure 2).

For the postural sway, Area ($F_{10,190}=2.53$; $p=0.007$) and RMS for both AP ($F_{10,190}=6.39$; $p<0.0001$) and ML ($F_{10,190}=3.19$; $p=0.001$) directions increased immediately after the fatigue protocol. Both the Area and RMS (AP and ML) decreased during the recovery phase after the fatigue protocol. The Area and the RMS-ML decreased and approached the baseline values after 4 min of recovery (Figure 3). Although the MSS showed the expected pattern (increase immediately after the fatigue protocol and gradual reduction during the recovery phase), no statistical difference was observed for either AP (baseline $[24.74\pm 7.68 \text{ mm.s}^{-1}]$ vs. immediately after the fatigue protocol $[31.21\pm 11.05 \text{ mm.s}^{-1}]$; $F_{10,190}=1.67$; $p=0.07$) or ML (baseline $[25.27\pm 6.28 \text{ mm.s}^{-1}]$ vs. immediately after the fatigue protocol $[28.75\pm 7.22 \text{ mm.s}^{-1}]$; $F_{10,190}=1.20$; $p=0.09$) directions. For the frequency analysis, the fatigue reduced the F80 in the ML direction ($F_{10,190}=2.08$; $p=0.027$), but there was no effect in the AP direction.

Figure 2. Mean and standard error of the mean for the heart rate (top), rate of perceived exertion (middle) and percentage of vertical force at the countermovement jump (bottom) at the baseline and every minute after the fatigue protocol.

* $p < 0.05$ ** $p < 0.0001$

Figure 3. Mean and standard error of the center of pressure (COP) mean for the area (top) and the root mean square in the Anteroposterior (middle) and mediolateral (bottom) directions at the baseline (Base.) and every minute after the fatigue protocol. * $p < 0.05$ ** $p < 0.0001$.

4. Discussion

The purpose of this study was to investigate the effect of a fatigue protocol specific for handball athletes in the F_{MAX} and the postural sway during quiet stance, acutely and during the first 10 min of recovery. The main results demonstrated that: i) the proposed protocol was effective to induce fatigue; ii) the loss of force persisted until the fourth minute; iii) postural sway increased after the fatigue protocol; and iv) COP Area and RMS-ML returned to baseline values at the fifth minute of the recovery phase.

4.1. The proposed protocol was effective to induce fatigue in female handball athletes

The proposed fatigue protocol was designed to cause high levels of stress and, consequently, exhaustion. The HR at the exhaustion (192.90 ± 4.45 bpm), which corresponds to $97.4 \pm 3.2\%$ of the predicted maximal HR, and the maximum values of the RPE scale (i.e., 10) demonstrated high cardiologic stress. Based on these values and the decrease of ~ 10 and 30% in the F_{MAX} and time, respectively, it is reasonable to consider that the proposed protocol can be used to induce fatigue in handball players.

The reduction in the ability to produce force relates to a decrease in the number of cross-bridges generating force and the number of cross-bridges working in the muscle (Nocella et al., 2011). However, fatigue has peripheral and central causes (Gandevia, 2001). The peripheral fatigue is characterized by a failure in the muscle to contract (Fitts, 1994). The central fatigue occurs with a considerable reduction of the neural drive to the muscles (Gandevia, 2001), which occurs after long-term efforts. Thus, it is plausible that the fatigue induced by our protocol has peripheral causes. However, according to the Central Governor Model (Pires, 2013), the central component can also contribute to this kind of exercise. This model considers that the Central Nervous System interacts with the muscle to protect important organs (e.g., brain and heart), decreasing the neural drive in high-demand situations, even during short-term exercises

(Noakes, 2000). The fatigue protocol of the present study may induce both peripheral and central fatigue.

4.2. The fatigue protocol impaired the postural control

Immediately after the fatigue protocol, the COP area and RMS increased. It is already known that fatigue increases postural sway (Boyas, McGown, Lajoie, & Bilodeau, 2013; Koyama & Yamauchi, 2017; Paillard, 2012; Pereira & Moraes, 2018; Thorlund et al., 2008; Zech et al., 2012). This increase results from alterations in different subsystems of the postural control system; however, how these subsystems are affected by the fatigue have to be further investigated. The somatosensory system is affected by internal alterations due to physical activities, and, in turn, the other systems are stimulated more than usual to supply the loss of inputs (Paillard, 2012). The fatigue alters the joint position sense (T.J. Allen & Proske, 2006; Forestier, Teasdale, & Nougier, 2002) and, consequently, can influence the postural sway. Moreover, during our fatigue protocol, eccentric and concentric muscle contractions were required, and they are associated with local glycogen depletion and metabolite accumulation. These alterations affect the effector system and change the proprioceptive response (Lepers, Bigard, Diard, Gouteyron, & Guezennec, 1997).

Although all of the participants were healthy, without any known vestibular or visual deficits, these subsystems could also be affected. A prolonged stimulus, as the fatigue protocol, decreased the vestibular sensitivity threshold (Lepers et al., 1997; Paillard, 2012). Also, during the fatigue protocol, the field of vision constantly changed, and the visual input was continuously stimulated to organize the body in different situations and tasks. The head was frequently oscillating and abruptly moving in the sequence of the tasks, which could give rise to a perception of body sway that would influence the postural compensation, even with somatosensory and vestibular system alterations (Lepers et al., 1997). Furthermore, the present

fatigue protocol increased both HR and RPE. Although we did not analyze the oxygen consumption, it is already known that there is a linear relationship between HR and the oxygen volume (Bot & Hollander, 2000). Paillard (2012) stated that the rise of voluntary ventilation decreased postural stability. More specifically, it could be that the fatigue of the inspiratory muscles would reduce the proprioceptive inputs from the low back, compromising the capacity of postural sway maintenance (Janssens, Brumagne, Polspoel, Troosters, & McConnell, 2010).

In summary, the fatigue induced by our protocol has the potential to cause changes in several systems involved with postural control. Some of these systems may be more affected than others, such as the somatosensory system. In addition, the other systems that generally try to overcome the somatosensory deficit were also likely affected by the fatigue protocol, increasing postural sway.

Despite this increase in postural sway, the COP speed showed no effect of fatigue. Different studies have shown the COP speed increased after fatigue (Beaulieu, Lamontagne, & Xu, 2008; Feuerbach & Grabiner, 1993; Pereira & Moraes, 2018). On the other hand, some studies did not find a difference in this postural control variable (Nardone, Tarantola, Giordano, & Schieppati, 1997), and others showed a reduction in COP speed (Koyama & Yamauchi, 2017). The high intensity of our protocol may have induced the fatigue of fast-twitch fibers of the gastrocnemius (i.e., powerful contractions), but not of the slow-twitch fibers of the soleus muscle (i.e., stability and postural control). During the balancing tasks, the soleus is more requested than the gastrocnemius, while during the power tasks (countermovement jump), the gastrocnemius activation is more relevant. These observations may help to explain the force decrease and the absence of effect in the COP speed (Koyama & Yamauchi, 2017).

4.3. The duration of the recovery phase depends on the variable analyzed

Different recovery time results from extrinsic (i.e., fatigue type protocol) and intrinsic

(i.e., muscle type fatigued, age, gender, and population) factors. The kind of fatigue protocol influences the intensity and duration of its effects on postural sway (Springer & Pincivero, 2009). Also, in a comparison between athletes and non-athletes on postural control, it has been found greater postural sway in non-athletes (Agostini, Chiaramello, Canavese, Bredariol, & Knaflitz, 2013; Baghbani, Woodhouse, & Gaeini, 2016), and quick recovery time in athletes (Ahmadi et al., 2017).

Different strategies are triggered to counteract or limit the effects of fatigue (Paillard, 2012). Although all systems are affected by the fatigue, the recovery time of each sensory system is different. During a general fatigue protocol, the vestibular system is affected, as well. However, it is less stimulated in comparison with the other systems and, consequently, a shorter time is needed to reset the receptors. Furthermore, visual adaptations were requested, providing inputs aiming to supply the somatosensory system. We asked the participants to look to a target positioned in front of them immediately after the fatigue protocol. A consequence of this sudden change in visual information is the recalibration of sensory integration. During the fatigue protocol, the visual system was highly requested to execute the task successfully. The visual system provides information about body orientation and sway, and postural compensations are observed (Lepers et al., 1997).

Also, as a result of the high-intensity exercise and fatigue, the hyperventilation affected postural control. We assumed a linear relationship between HR and oxygen uptake. This decrease helped in the postural re-adaptation. Based on the model that during the recovery phase, the blood circulation and oxygen consumption are reduced, it is possible the increase of neural drive for distal organs. Further, the reduction in the hyperventilation during the recovery time decreased the effects in the peripheral nerve fibers (Sakellari & Bronstein, 1997), and proprioceptive input from the lower back may be more reliable and help in the postural maintenance (Janssens et al., 2010).

Despite the fast recovery observed for some COP parameters, the AP RMS did never return to baseline values within 10 minutes. This finding may result from the postural task used and the fatigued muscles. Feuerbach and Grabiner (1993) found higher values for COP AP displacement during a single-leg task. Similar results were found by Tanaka et al. (1996), where the participants showed higher values in the AP direction after the fatigue of the ankle muscles. These muscles are responsible for the ankle joint stabilization and, consequently, by the ankle strategy. However, when fatigued, these muscles lost their ability to produce force and the ability to transmit information through local nerves (Koyama & Yamauchi, 2017). With the decrease in force, the COP RMS in the AP direction is more affected than in the ML direction. When the plantar flexor and dorsiflexor muscles are affected, the impairments in postural control are more evident in the AP direction (Gribble & Hertel, 2004).

4.4. Limitations

A potential limitation of this study is the sampling duration of the postural task that can influence some COP variables (e.g., RMS and speed (Carpenter, Frank, Winter, & Peysar, 2001)). Our option to collect 30-s of COP data was because we wanted to collect the force data at every minute as well. For the sake of certainty, it is suggested that futures investigations should use a longer sampling duration (i.e., 60 s) to confirm the present findings. Another potential limitation is the impossibility to differentiate between central and peripheral causes of fatigue. Thus, future studies should consider incorporating methods to identify the causes of fatigue in this kind of protocol. Lastly, aiming at an ecological approach, future investigations should include single-leg jumps in the fatigue protocol and also analyze the force and postural sway patterns during a real game situation.

5. Conclusion

The fatigue protocol was efficient to induce fatigue in female handball players. Besides, the handball players increased their postural sway and decreased their capacity to produce force during the countermovement jump in the fatigue situation. However, the force returned to baseline values after five minutes of the fatigue protocol. Also, some postural sway variables returned to baseline values at the fifth minute of the recovery time while others persisted during a more extended period. Therefore, future studies willing to investigate the effects of fatigue in female handball athletes should perform the analysis during this short time window.

Declaration of interest statement

None declared.

Funding

This work was supported by the Coordenação de Aperfeiçoamento de Pessoal de Nível Superior - Brasil (CAPES) - Finance Code 001 and by São Paulo Research Foundation – FAPESP (grant #2018/20362-8).

References

- [1] S.C. Gandevia, Spinal and supraspinal factors in human muscle fatigue, *Physiological reviews* 81(4) (2001) 1725-89.
- [2] J.D. Chappell, D.C. Herman, B.S. Knight, D.T. Kirkendall, W.E. Garrett, B. Yu, Effect of fatigue on knee kinetics and kinematics in stop-jump tasks, *Am J Sports Med* 33(7) (2005) 1022-9.
- [3] L.A. Hiemstra, I.K. Lo, P.J. Fowler, Effect of fatigue on knee proprioception: implications for dynamic stabilization, *The Journal of orthopaedic and sports physical therapy* 31(10) (2001) 598-605.
- [4] S. Steib, A. Zech, C. Hentschke, K. Pfeifer, Fatigue-induced alterations of static and dynamic postural control in athletes with a history of ankle sprain, *J Athl Train* 48(2) (2013) 203-8.
- [5] S. Dieling, M. van der Esch, T.W. Janssen, Knee joint proprioception in ballet dancers and non-dancers, *Journal of dance medicine & science : official publication of the International Association for Dance Medicine & Science* 18(4) (2014) 143-8.
- [6] F.A. Barbieri, S.T. Rodrigues, P.F. Polastri, R.A. Barbieri, P.H.A. de Paula, F. Milioni, et al., High intensity repeated sprints impair postural control, but with no effects on free throwing accuracy, in under-19 basketball players, *Human movement science* 54 (2017) 191-196.
- [7] M. Ahmadi, A. Nodehi Moghadam, M. Bakhshi, T. Rezaeian, G. Pahnabi, The duration effects of lumbar extensor muscle fatigue on postural control in soccer players, *Medical journal of the Islamic Republic of Iran* 31 (2017) 112.
- [8] P.A. Gribble, J. Hertel, Effect of lower-extremity muscle fatigue on postural control, *Archives of physical medicine and rehabilitation* 85(4) (2004) 589-92.
- [9] A. Zech, S. Steib, C. Hentschke, H. Eckhardt, K. Pfeifer, Effects of localized and general fatigue on static and dynamic postural control in male team handball athletes, *Journal of strength and conditioning research* 26(4) (2012) 1162-8.
- [10] Z.G. Fox, J.P. Mihalik, J.T. Blackburn, C.L. Battaglini, K.M. Guskiewicz, Return of postural control to baseline after anaerobic and aerobic exercise protocols, *J Athl Train* 43(5) (2008) 456-63.
- [11] L. Guidetti, E. Franciosi, M.C. Gallotta, G.P. Emerenziani, C. Baldari, Postural control after a prolonged treadmill run at individual ventilatory and anaerobic threshold, *Journal of sports science & medicine* 10(3) (2011) 515-9.
- [12] D. Dickin, J. Doan, Postural stability in altered and unaltered sensory environments following fatiguing exercise of lower extremity joints, *Scandinavian journal of medicine & science in sports* 18(6) (2008) 765-772.
- [13] P.A. Gribble, J. Hertel, C.R. Denegar, W.E. Buckley, The Effects of Fatigue and Chronic Ankle Instability on Dynamic Postural Control, *J Athl Train* 39(4) (2004) 321-329.

- [14] K.A. Webster, B.G. Pietrosimone, P.A. Gribble, Muscle Activation During Landing Before and After Fatigue in Individuals With or Without Chronic Ankle Instability, *J Athl Train* (2016).
- [15] J.B. Thorlund, L.B. Michalsik, K. Madsen, P. Aagaard, Acute fatigue-induced changes in muscle mechanical properties and neuromuscular activity in elite handball players following a handball match, *Scand J Med Sci Sports* 18(4) (2008) 462-72.
- [16] T.E. Hewett, K.R. Ford, G.D. Myer, Anterior cruciate ligament injuries in female athletes - Part 2, a meta-analysis of neuromuscular interventions aimed at injury prevention, *Am J Sport Med* 34(3) (2005) 490-498. <Go to ISI>://WOS:000235449900020.
- [17] G. Langevoort, G. Myklebust, J. Dvorak, A. Junge, Handball injuries during major international tournaments, *Scandinavian journal of medicine & science in sports* 17(4) (2007) 400-407. <Go to ISI>://WOS:000248481300015.
- [18] M.J. Stuart, A. Smith, Injuries in Junior A ice hockey. A three-year prospective study, *Am J Sports Med* 23(4) (1995) 458-61.
- [19] C. Foster, J.A. Florhaug, J. Franklin, L. Gottschall, L.A. Hrovatin, S. Parker, et al., A new approach to monitoring exercise training, *Journal of strength and conditioning research* 15(1) (2001) 109-15.
- [20] M. Duarte, S.M.S.F. Freitas, Revisão sobre posturografia baseada em plataforma de força para avaliação do equilíbrio, *Brazilian Journal of Physical Therapy* 14 (2010) 183-192. http://www.scielo.br/scielo.php?script=sci_arttext&pid=S1413-35552010000300003&nrm=iso.
- [21] R. Moraes, B.L.S. Bedo, L.O. Santos, R.A. Batistela, P.R.P. Santiago, E. Mauerberg-deCastro, Additional Haptic Information Provided by Anchors Reduces Postural Sway in Young Adults Less Than Does Light Touch, *Frontiers in neuroscience* 12 (2018) 346.
- [22] M. Nocella, B. Colombini, G. Benelli, G. Cecchi, M.A. Bagni, J. Bruton, Force decline during fatigue is due to both a decrease in the force per individual cross-bridge and the number of cross-bridges, *The Journal of physiology* 589(Pt 13) (2011) 3371-3381. <https://www.ncbi.nlm.nih.gov/pubmed/21540343>
<https://www.ncbi.nlm.nih.gov/pmc/PMC3145945/>.
- [23] R.H. Fitts, Cellular mechanisms of muscle fatigue, *Physiological reviews* 74(1) (1994) 49-94.
- [24] F.O. Pires, Thomas Kuhn's 'Structure of Scientific Revolutions' applied to exercise science paradigm shifts: example including the Central Governor Model, *Br J Sports Med* 47(11) (2013) 721-2.
- [25] T.D. Noakes, Physiological models to understand exercise fatigue and the adaptations that predict or enhance athletic performance, *Scandinavian journal of medicine & science in sports* 10(3) (2000) 123-45.

- [26] S. Boyas, C. McGown, Y. Lajoie, M. Bilodeau, Changes in posture associated with postural control impairment induced by ankle plantarflexor fatigue during unipedal quiet standing, *Neuroscience letters* 547 (2013) 48-52.
- [27] T. Paillard, Effects of general and local fatigue on postural control: a review, *Neuroscience and biobehavioral reviews* 36(1) (2012) 162-76.
- [28] A.B.R. Pereira, R. Moraes, The Use of the Anchor System Reduces Postural Sway During Upright Standing Irrespective of Plantar Flexors Muscle Fatigue in Young and Older Adults, *Motor control* 22(3) (2018) 338-357.
- [29] K. Koyama, J. Yamauchi, Altered postural sway following fatiguing foot muscle exercises, *PLoS One* 12(12) (2017) e0189184.
- [30] T.J. Allen, U. Proske, Effect of muscle fatigue on the sense of limb position and movement, *Experimental Brain Research* 170(1) (2006) 30-38. <https://doi.org/10.1007/s00221-005-0174-z>.
- [31] N. Forestier, N. Teasdale, V. Nougier, Alteration of the position sense at the ankle induced by muscular fatigue in humans, *Med Sci Sports Exerc* 34(1) (2002) 117-22.
- [32] R. Lepers, A.X. Bigard, J.-P. Diard, J.-F. Gouteyron, C.Y. Guezennec, Posture control after prolonged exercise, *European journal of applied physiology and occupational physiology* 76(1) (1997) 55-61. <http://dx.doi.org/10.1007/s004210050212>.
- [33] S.D. Bot, A.P. Hollander, The relationship between heart rate and oxygen uptake during non-steady state exercise, *Ergonomics* 43(10) (2000) 1578-92.
- [34] L. Janssens, S. Brumagne, K. Polspoel, T. Troosters, A. McConnell, The effect of inspiratory muscles fatigue on postural control in people with and without recurrent low back pain, *Spine* 35(10) (2010) 1088-94.
- [35] J.W. Feuerbach, M.D. Grabiner, Effect of the aircast on unilateral postural control: amplitude and frequency variables, *The Journal of orthopaedic and sports physical therapy* 17(3) (1993) 149-54.
- [36] M.L. Beaulieu, M. Lamontagne, L. Xu, Gender differences in time-frequency EMG analysis of unanticipated cutting maneuvers, *Med Sci Sports Exerc* 40(10) (2008) 1795-804.
- [37] A. Nardone, J. Tarantola, A. Giordano, M. Schieppati, Fatigue effects on body balance, *Electroencephalography and clinical neurophysiology* 105(4) (1997) 309-20.
- [38] B.K. Springer, D.M. Pincivero, The effects of localized muscle and whole-body fatigue on single-leg balance between healthy men and women, *Gait & posture* 30(1) (2009) 50-4.
- [39] V. Agostini, E. Chiaramello, L. Canavese, C. Bredariol, M. Knaflitz, Postural sway in volleyball players, *Human movement science* 32(3) (2013) 445-56.
- [40] F. Baghbani, L.J. Woodhouse, A.A. Gaeini, Dynamic Postural Control in Female Athletes and Nonathletes After a Whole-Body Fatigue Protocol, *Journal of strength and conditioning research* 30(7) (2016) 1942-7.

- [41] V. Sakellari, A.M. Bronstein, Hyperventilation effect on postural sway, Archives of physical medicine and rehabilitation 78(7) (1997) 730-6.
- [42] T. Tanaka, N. Hashimoto, M. Nakata, T. Ito, S. Ino, T. Ifukube, Analysis of toe pressures under the foot while dynamic standing on one foot in healthy subjects, The Journal of orthopaedic and sports physical therapy 23(3) (1996) 188-93.
- [43] M.G. Carpenter, J.S. Frank, D.A. Winter, G.W. Peysar, Sampling duration effects on centre of pressure summary measures, Gait & posture 13(1) (2001) 35-40.

C APÍTULO 3

ESTUDO 2

The effect of fatigue on hip and knee kinematics during single- and double-leg dynamic tasks: A study on female handball players

Bruno Luiz Souza Bedo; Danilo S. Catelli; Renato Moraes; Mario Lamontagne; Dayanne Rodrigues Pereira; Julia Barachini Graça; Paulo Roberto Pereira Santiago.

The effect of fatigue on hip and knee kinematics during single- and double-leg dynamic tasks: A study on female handball players

Bruno Luiz Souza Bedo^{A,B,C}; Danilo S. Catelli^C; Renato Moraes^{A,B}; Mario Lamontagne^C; Dayanne Rodrigues Pereira^B; Julia Barachini Graça^A; Paulo Roberto Pereira Santiago^{A,B}

^ASchool of Physical Education and Sports of Ribeirão Preto, University of São Paulo, Ribeirão Preto, Brazil;

^BRibeirão Preto Medical School, University of São Paulo, Ribeirão Preto, Brazil;

^CSchool of Human Kinetics, University of Ottawa, Ottawa, Canada.

Observação:

O presente estudo encontra-se em processo de revisão.

Abstract

This study investigated the effect of a handball-specific fatigue protocol in the hip and knee kinematics. Twenty female handball athletes performed three trials of the single-leg landing (SLL), sidestep cutting maneuver (SCM), and drop vertical jump (DVJ) before and after the fatigue protocol. Knee and hip angles waveforms were compared using the statistical parametric mapping, with $p < 0.05$. During the SLL, the fatigue increased hip adduction (4-7% cycle, $p = 0.045$) and knee abduction (4-9%, $p = 0.037$; 25-27%, $p = 0.048$). For the SCM, the fatigue reduced the hip flexion during 14-29% and 44-68% of the cycle ($p = 0.031$ and 0.014 , respectively). Similarly, the knee flexion decreased between 7-36% and 53-73% ($p = 0.0001$ and 0.0017 , respectively). Besides, during the fatigue state, the athletes reduced the hip abduction between 0-11% of the cycle ($p = 0.043$) and increased the knee abduction between 20-23% ($p = 0.043$). During the DVJ task, the fatigue decreased the hip flexion between 19-44% of the cycle ($p = 0.002$) and the knee flexion between 1-16% and 18-77%. The fatigue protocol altered the lower limb kinematics, decreasing the knee and hip flexion during the SCM and DVJ, and increasing the knee valgus during both single-leg landing tasks.

Keywords: Handball; Fatigue; Knee; Kinematics; sidestep cutting maneuver.

Introduction

Female athletes have a higher risk of non-contact anterior cruciate ligament (ACL) rupture compared to male athletes (Ford, Myer, & Hewett, 2003; Harmon & Ireland, 2000), especially in modalities involving rapid changes of direction, jumping/landing and pivoting, as it is the case in handball (Bedo et al., 2019; Olsen et al., 2004, 2006). Moreover, the ACL rupture usually occurs during single- and double-leg landings or sidestep cutting maneuvers with a slight knee flexion associated with knee valgus and internal rotation (Beaulieu, Lamontagne, & Xu, 2009; Hewett, Torg, & Boden, 2009). Due to a large number of athletic injuries observed during these tasks, there has been a growing interest in investigating the biomechanics of such tasks in different laboratory settings (Besier, Lloyd, Cochrane, & Ackland, 2001; Cesar, Havens, Chang, & Sigward, 2011; Kristianslund, Faul, Bahr, Myklebust, & Krosshaug, 2014; McLean, Huang, & van den Bogert, 2005; McLean, Lipfert, & van den Bogert, 2004; Sigward, Cesar, & Havens, 2015)

The gender difference for the risk of injury is a consequence of anatomic, environmental, hormonal, and biomechanical aspects ("The female ACL: Why is it more prone to injury?," 2016), and researchers have investigated the neuromuscular aspects as one factor that improve the understanding of the injury mechanism (Beaulieu et al., 2008; Norcross, Blackburn, Goerger, & Padua, 2010; Sigward & Powers, 2007). Previous studies have related muscle fatigue to increased risk of ACL injury (Borotikar et al., 2008; Liederbach et al., 2014; McLean et al., 2007; Xia et al., 2017) and recent systematic Review pointed out that fatigue has no significant influence on most of the knee kinematic and kinetic variables (Benjaminse et al., 2019), and few prospective studies have quantified suitable parameters to estimate the effects of fatigue in the ACL injury risk (Bourne et al., 2019). Therefore, the understanding of the potential mechanism by which the fatigue manifests in the injury remains unclear (McLean et al., 2007), particularly the risk of ACL injury (Benjaminse et al., 2019; Bourne et al., 2019;

McLean et al., 2007). A possible limiting factor of those previous studies investigating the effect of fatigue is the lack of fatigue protocols that would consistently affect lower limb kinematics since those fatigue protocols seem to oversimplify the complexity related to fatigue (Benjaminse et al., 2019).

Fatigue is defined as a reduction in the ability to exert muscle force or power (Gandevia, 2001). Different fatigue protocols have used a treadmill (Benjaminse et al., 2008; Zech et al., 2012), step-up and bounding tasks (McLean et al., 2007), isokinetic dynamometer (Gribble et al., 2004), single-leg elevation (Zech et al., 2012), and small games (Thorlund et al., 2008). Nevertheless, general fatigue protocol has seemed to better investigate the links between lower limb neuromuscular control and non-contact ACL injury risk (McLean et al., 2005), and the connection between fatigue and decision making may represent the worst scenario to risk of knee injuries (Borotikar et al., 2008). But to our knowledge, motor skills used in a handball game have not been used in fatigue protocols and no published fatigue protocols are showing a consistent effect on lower limb kinematics to increase the ACL injury risk (Bourne et al., 2019). However, a recent investigation has presented a specific fatigue protocol composed of handball actions that simulate the handball movements (Bedo, Pereira, et al., 2020).

Furthermore, to elaborate effective preventive injury programs in the future, it is essential to approximate the laboratory environment to the field to increase the ecological validity of the study, as a sport-realistic protocol provides a better understanding of injury mechanisms related to the knee joint during the execution of dynamic tasks (Besier, Lloyd, Ackland, & Cochrane, 2001). Therefore, the purpose of this study was to investigate the effect of a handball-specific fatigue protocol in the hip and knee kinematics in female handball players during the single- and double-leg dynamic tasks. We hypothesized that compared with the baseline values, the hip and knee kinematics would change after the fatigue protocol.

Methods

Participants

Twenty female handball athletes (21.9 ± 3.4 years; 63.5 ± 9.1 kg; 1.76 ± 0.07 m; 7.2 ± 3.2 years of experience playing handball; 240 ± 180.4 minutes of practice/week) with no report of ACL rupture and without any other lower limb injury in the last six months participated in this study. All participants were right-handed, and the lower limb dominance was determined by asking the leg used to take off in a handball jump shot (Zech et al., 2012). The local Human Research Ethics Committee granted ethical approval, and all participants signed a consent form before data collection.

Fatigue protocol

The fatigue protocol was performed in the laboratory environment in a circuit format focused on movement performed during handball practice. The protocol was elaborated based in movements performed during handball practice, sidestep cutting, jumping, landing, lateral and back shifting, sprints, among others (see Bedo, Pereira, et al. (2020) for details). The athletes performed two laps in the circuit for acclimation and warm-up. The protocol was divided into rounds with a gradual increase in volume, where the number of laps in the circuit was the number of rounds (i.e., round I [1 lap]; round II [2 laps]; and so on). The presence of fatigue was based on self-report of exhaustion, characterized by the participant's inability to continue performing the protocol.

Experimental procedure

The fatigue parameters, such as heart rate (HR, Polar V800TM, FIN) and the rate of perceived exertion (using the Foster scale [0-10]), and the time of effort were assessed priorly and after the fatigue protocol. In addition, the maximum force was quantified by a

countermovement jump prior and immediately after the participant's exhaustion to quantify the perceptual reduction in force production. During the countermovement jump the participants were instructed to jump as high as possible without flexing the knees and hips during the flight phase

Figure 1. The schematic representation of the data collection. For the recording of the maximum force propulsion, a force plate was used during the countermovement jump (A and D). The participants performed three valid trials of each task before (B) and after (E) the fatigue protocol, which was based on motor skills performed during handball practice (C). The yellow leg indicates the limb evaluated.

The athletes performed three trials of each task in the following sequence: single-leg landing (SLL), sidestep cutting maneuver (SCM), and drop vertical jump (DJV), before and after the fatigue protocol (Figure 1). In the SLL task, the participants stepped down from a 40-cm box onto the floor. They were instructed to stepping off the box and land on the non-dominant leg, as comfortably and normally as possible without falling over or touching the ground with either their hands or dominant leg. A successful trial occurred when the participant stepped off the box (without an upward and/or forward jump action) and adopted a stable landing posture on the ground without losing balance for two seconds. In the SCM task, the athletes sprinted for four meters as fast as possible and performed a 45° cutting maneuver from the original movement with the non-dominant leg. The athletes continued running after sidestepping execution for approximately three meters. A successful trial occurred when the participant remained over the color tape placed on the floor, used to demarcate the direction of the cutting maneuver. The DJV was also performed on the 40-cm high box. Participants were instructed to “*drop off from the front of the box, land with their feet together, and immediately jump as high as possible*”. A successful trial occurred when the athlete landed with their feet together on the floor and adopted a stable landing after the jump. All participants wore their handball shoes. The tasks were performed with a one-minute interval, during the first five minutes after exhaustion, while the fatigue effects were still present (Bedo, Pereira, et al., 2020).

Instrumentation and data processing

To quantify the perceptual change in maximum vertical jump force production (F_{MAX}), the athletes performed a countermovement jump over a force plate (40 × 60 cm, Bertec, Columbus, USA) before and immediately after the fatigue protocol. Data from the force plate were collected using an A/D converter (National Instruments #USB-6251-BNC, Austin, TX,

USA) sampled at 5,000 Hz. The data were collected and processed on Matlab® (Mathworks Inc. Natick, MA, USA), filtered with a low-pass fourth-order Butterworth filter with a 10-Hz cut-off frequency. The F_{MAX} was determined as the first vertical peak of the vertical ground reaction force, and normalized by the participants' total body weight. The percentage of F_{MAX} alteration was quantified between the baseline and the fatigue values.

To calculate the joint angles, an eight-camera motion capture system (MX-T40S, Vicon, Oxford, UK) tracked the reflective markers at 250 Hz. The University of Ottawa Motion Analysis Model (UOMAM) marker set was used (Mantovani & Lamontagne, 2017a). The marker trajectories were reconstructed, labeled, and filtered (zero-lag, fourth-order Butterworth filter, cut-off frequency = 6 Hz) in Nexus 2.8 (Vicon, Oxford, UK). The hip and knee angles were calculated based on Euler angles rotation for a 2-s interval after the initial ground contact for the SLL and during the stance phase (between foot-strike and foot-off) for both SCM and DVJ tasks.

Statistical analysis

The normality and sphericity assumptions of the data for the fatigue variables (heart rate, rate of perceived exertion and F_{MAX}), and the time of effort of the first and last laps were confirmed using Shapiro-Wilk and Mauchly's tests, respectively. These variables were compared using paired samples t-tests using the Statistical Package for the Social Sciences (SPSS Statistics v.23, IBM Corporation, Armonk, USA).

For the comparison of the effect of fatigue in both hip and knee kinematics during the SLL, SCM, and DVJ, the angles were time-normalized (0-100%) and the waveforms of sagittal, frontal and transverse planes were compared by paired analysis using statistical parametric mapping (SPM) (Pataky, 2010) with the significance level set at $p < 0.05$.

Results

The heart rate (baseline: 82.45 ± 8.14 bpm *vs.* fatigue: 194.35 ± 7.38 bpm) and the rate of perceived exertion (baseline: 0 ± 0 *vs.* fatigue: 10 ± 0) and the effort time (first lap = 42.12 ± 4.95 seconds *vs.* last lap: 58.85 ± 9.39 seconds) increased significantly after the fatigue protocol (all variables $p < 0.0001$). The F_{MAX} reduced $\sim 10\%$ immediately after the fatigue protocol compared with baseline values (baseline: 2.51 ± 0.51 BW *vs.* fatigue: 2.24 ± 0.30 BW; $p < 0.0001$).

The presence of fatigue influenced the knee and hip kinematics. During SLL, the SPM analysis showed an increase in hip adduction in the fatigue condition immediately after ground contact (between 4-7% of the cycle, $p = 0.045$). For the knee joint, there was a larger knee abduction during the fatigue state compared to baseline, and also after ground contact (between 4-9%, $p = 0.037$, and 25-27%, $p = 0.48$, of the SLL cycle). There were no kinematic differences in the sagittal and transversal planes (Figure 2).

Concerning the SCM task, during the fatigue state, a significant reduction in hip flexion was detected during the 14-29% and 44-68% of the SCM cycle ($p = 0.031$ and 0.014 , respectively). Similarly, there was a decrease in knee flexion at 7-36% and 53-73% of the cycle ($p = 0.0001$ and 0.0017 , respectively). In addition, the fatigue reduced hip abduction between 0-11% of the cycle ($p = 0.43$), whereas it increased knee abduction between 20-23% of the cycle ($p = 0.043$). There was no effect of fatigue in the transverse plane (Figure 3).

Figure 2. Hip (top) and knee (bottom) joint angles during the single-leg landing (SLL) task. The left, middle, and right plots represent the sagittal, frontal, and transversal planes, respectively. The red and black lines represent the average angles for the fatigue and baseline conditions, respectively. The corresponding colors of the shaded areas represent the standard deviation of each condition. The red bars with black mosaicking denote the significant differences areas indicated by the SPM analysis. The illustration in the bottom represents the time-normalized movement window investigated during the SLL task. The yellow leg represents the assessed limb.

Figure 3. Hip (top) and knee (bottom) joint angles during the sidestep cutting maneuver (SCM) task. The left, middle, and right plots represent the sagittal, frontal, and transversal planes, respectively. The red and black lines represent the average angles for the fatigue and baseline conditions, respectively. The corresponding colors of the shaded areas represent the standard deviation of each condition. The red bars with black mosaicking denote the significant differences areas indicated by the SPM analysis. The illustration in the bottom represents the time-normalized movement window investigated during the SCM task. The yellow leg represents the assessed limb.

During the DVJ task, the fatigue affected only the sagittal plane for both knee and hip joints. The fatigue reduced the hip flexion during the 19-44% of the cycle ($p = 0.002$). Similarly, there was a decrease in knee flexion during fatigue at 1-16% and 18-77% of the cycle. The fatigue did not affect the joint angles in both frontal and transversal planes (Figure 4).

Figure 4. Hip (top) and knee (bottom) joint angles during the drop vertical jump (DVJ) task. The left, middle, and right plots represent the sagittal, frontal, and transversal planes, respectively. The red and black lines represent the average angles for the fatigue and baseline conditions, respectively. The corresponding colors of the shaded areas represent the standard deviation of each condition. The red bars with black mosaicking denote the significant differences areas indicated by the SPM analysis. The illustration in the bottom represents the time-normalized movement window investigated during the DVJ task. The yellow leg represents the assessed limb.

Discussion

We investigated the effect of a specific fatigue protocol for handball athletes in hip and knee kinematics during single- and double-leg dynamic tasks. The proposed hypothesis was that the hip and knee kinematics would change after the fatigue protocol. The current protocol

altered the postural control in handball athletes (Bedo, Pereira, et al., 2020), and, in the present study, it also supported the premise that fatigue alters the lower limb kinematics (Borotikar et al., 2008). More specifically, we observed an increase in knee valgus during the single-leg tasks (SLL and SCM) and a reduction in knee and hip flexion during both SCM and DVJ tasks following the fatigue protocol. Moreover, with fatigue, we found an increase in hip adduction in the SCM task and an increase in hip adduction for the SLL task. These results provide insights into how fatigue affects knee kinematics helping to prevent injuries.

The majority of these changes occurred during the weight-acceptance period (immediately following the initial ground contact), although a few changes persisted until the middle of the movement cycle, as observed in the double-leg tasks. The instants immediately following ground contact have been showing to be crucial to the occurrence of a knee injury, such as ACL rupture (Koga et al., 2010), mainly when the knee joint is close to full extension (Hewett et al., 2009; Koga et al., 2010; Makaruk, Czaplicki, Sacewicz, & Sadowski, 2014; Olsen et al., 2004; Wang, 2011). The reduction in knee flexion may contribute to knee injuries, since it relates to the drawer mechanism, increasing the shear component during small knee flexion angles (Draganich, Andriacchi, & Andersson, 1987; Koga et al., 2010; Laughlin et al., 2011). In the present study, fatigue reduced knee flexion in both SCM and DVJ tasks. Although we did not evaluate the muscle activation, our fatigue protocol certainly affected the lower limb muscles, as the F_{MAX} decreased after the fatigue protocol. O'Connor, Johnson, and Benson (2015) fatigued the hamstrings and noticed a reduction in the knee flexion while performing dynamic landing. Moreover, the reduced capacity of the hamstrings to produce force likely affected the knee load, since a lower knee flexion increases the patella tendon-tibia shaft angle, increasing the anterior shear force at the proximal end of the tibia (Nunley, Wright, Renner, Yu, & Garrett Jr, 2003; B. Yu & Garrett, 2007).

In addition, there was an increase in knee valgus for the SLL and SCM tasks during the weight-acceptance phase in the fatigue condition. The increase in knee valgus has also been related to the risk of ACL injuries (Besier, Lloyd, Cochrane, et al., 2001; Boden, Dean, Feagin, & Garrett, 2000; Herrington, 2011; Hewett et al., 2005). When the athletes performed the task with an increase in the knee valgus, the medial collateral ligament becomes taut, and lateral compression occurs. The anterior force vector caused by quadriceps contraction causes a displacement of the femur relative to the tibia, resulting in the risk of ACL injury (Koga et al., 2010). The increase in knee valgus may result from fatigue affecting the mechanoreceptor feedback, which impacts joint stability (Hiemstra et al., 2001; Lattanzio, Petrella, Sproule, & Fowler, 1997). Moreover, fatigue also affects the Golgi tendon organs response (Gandevia, 2001), impacting joint stability, especially in women, who usually have less muscular protection of the ligaments during the external loading of the knee (Wojtys, Huston, Schock, Boylan, & Ashton-Miller, 2003).

The differences in the frontal plane were found only in the single-leg tasks. Differently, in the sagittal plane, lower values of knee and hip flexion were found when the athletes were fatigued. The differences may be dependent on the nature of these tasks. Single-leg tasks are functionally more demanding since one leg is loaded while the center of mass decelerates in both vertical and horizontal directions (Orishimo & Kremenec, 2006). These findings corroborate with previous studies that have shown a higher risk of knee injury in single- than in double-leg landing tasks (Wang, 2011). However, although the knee is the most injured, the hip is the primary joint used to support and stabilize the landing during the absorption phase (Decker, Torry, Wyland, Sterett, & Richard Steadman, 2003). Thus, the decrease in hip flexion may result from the inability of hip and knee extensors to eccentrically control the downward progression during the absorption phase (Borotikar et al., 2008).

The relationship between hip adduction and knee abduction has already been described as an important aspect that increases the risk of ACL injury (Hewett & Myer, 2011). Thus, these alterations during the fatigue state likely increase the risk of ACL injury in female athletes. However, the exact mechanisms by which fatigue affects the knee and hip joint motion remain unclear, and there appear to have several explanations (Borotikar et al., 2008), such as substantial movement control adaptations (Nyland, Caborn, Shapiro, & Johnson, 1997), mechanoreceptor dysfunction (Miura et al., 2004), increase in the joint laxity (Wojtys et al., 1996), and both peripheral and central causes (Gandevia, 2001).

Although peripheral fatigue cannot be avoided, it is part of sports practice, and, consequently, it would contribute to increasing the risk of ACL injury. Differently, the increase in the complexity of the exercises, challenging decision-making tasks, and neuromuscular training likely represent possible strategies to counter the deleterious impact of fatigue on central processing (McLean et al., 2007). Besides, it is necessary to consider how fatigue affects the lower limb joints during realistic sports situations (i.e., current protocol). General protocols appear to more practically simulate sports relevant movement tasks and may induce both local and central fatigue effects, impacting the proprioceptive system (T.J. Allen & Proske, 2006; Hiemstra et al., 2001; Miura et al., 2004).

There are potential limitations in this study that can influence the statement that fatigue impact the knee kinematics of handball players. First, this study was performed in a laboratory setting, and, although we used a fatigue protocol and motor skills similar to the ones performed during the handball practices, aiming at an ecological validity, future studies should be conducted on the court using an actual game to induce fatigue. Second, we investigated only kinematic variables. Future investigations should include kinetic parameters and muscle activity to quantify the fatigue effects on muscle forces and joint contact forces. . Lastly, future studies should include methods to identify the causes of fatigue (central or peripheral) in this

kind of protocol. Finally, aiming at an ecological approach, future investigations should include dynamic tasks during a real game.

Conclusion

The fatigue altered the lower limb kinematics, decreasing the knee and hip flexion during the SCM and DVJ, and increasing the knee valgus during both single-leg landing tasks (SCM and SLL), likely increasing the risk of an ACL injury.

Acknowledgments

The authors thank the technical staff and athletes for participating and contributing to the current study.

Funding

This study was financed in part by the Coordenação de Aperfeiçoamento de Pessoal de Nível Superior – Brasil (CAPES) – Finance Code 001; by the São Paulo Research Foundation – FAPESP, Brazil [grants 2018/20362-8 and 2019/04721-0]; and also had a financial contribution from the Human Movement Biomechanics Laboratory and from the Brazilian Society of Biomechanics.

References

- Allen, T. J., & Proske, U. (2006). Effect of muscle fatigue on the sense of limb position and movement [journal article]. *Experimental brain research*, 170(1), 30-38. <https://doi.org/10.1007/s00221-005-0174-z>
- Beaulieu, M. L., Lamontagne, M., & Xu, L. (2008). Gender differences in time-frequency EMG analysis of unanticipated cutting maneuvers. *Med Sci Sports Exerc*, 40(10), 1795-1804.
- Beaulieu, M. L., Lamontagne, M., & Xu, L. (2009). Lower limb muscle activity and kinematics of an unanticipated cutting manoeuvre: a gender comparison. *Knee Surg Sports Traumatol Arthrosc*, 17(8), 968-976. <https://doi.org/10.1007/s00167-009-0821-1>
- Bedo, B. L. S., Manechini, J. P. V., Nunomura, M., Menezes, R. P., & Silva, S. R. D. (2019). Injury Frequency in Handball Players: A Descriptive Study of Injury Pattern in São Paulo State Regional Teams. *Motriz: Revista de Educação Física*.
- Bedo, B. L. S., Pereira, D. R., Moraes, R., Kalva-Filho, C. A., Will-de-Lemos, T., & Santiago, P. R. P. (2020). The rapid recovery of vertical force propulsion production and postural sway after a specific fatigue protocol in female handball athletes. *Gait Posture*, 77, 52-58. <https://doi.org/https://doi.org/10.1016/j.gaitpost.2020.01.017>
- Benjaminse, A., Habu, A., Sell, T. C., Abt, J. P., Fu, F. H., Myers, J. B., & Lephart, S. M. (2008). Fatigue alters lower extremity kinematics during a single-leg stop-jump task. *Knee Surg Sports Traumatol Arthrosc*, 16(4), 400-407. <https://doi.org/10.1007/s00167-007-0432-7>
- Benjaminse, A., Webster, K. E., Kimp, A., Meijer, M., & Gokeler, A. (2019). Revised Approach to the Role of Fatigue in Anterior Cruciate Ligament Injury Prevention: A Systematic Review with Meta-Analyses. *Sports Med*, 49(4), 565-586. <https://doi.org/10.1007/s40279-019-01052-6>
- Besier, T. F., Lloyd, D. G., Ackland, T. R., & Cochrane, J. L. (2001). Anticipatory effects on knee joint loading during running and cutting maneuvers. *Med Sci Sports Exerc*, 33(7), 1176-1181.
- Besier, T. F., Lloyd, D. G., Cochrane, J. L., & Ackland, T. R. (2001). External loading of the knee joint during running and cutting maneuvers. *Medicine and Science in Sports and Exercise*, 33(7), 1168-1175. <Go to ISI>://WOS:000169660600014
- Boden, B. P., Dean, G. S., Feagin, J. A., & Garrett, W. E. (2000). Mechanisms of anterior cruciate ligament injury. *Orthopedics*, 23(6), 573-578. <Go to ISI>://WOS:000087656800010
- Borotikar, B. S., Newcomer, R., Koppes, R., & McLean, S. G. (2008). Combined effects of fatigue and decision making on female lower limb landing postures: central and peripheral contributions to ACL injury risk. *Clin Biomech (Bristol, Avon)*, 23(1), 81-92. <https://doi.org/10.1016/j.clinbiomech.2007.08.008>

- Bourne, M. N., Webster, K. E., & Hewett, T. E. (2019). Is Fatigue a Risk Factor for Anterior Cruciate Ligament Rupture? *Sports Med*, 49(11), 1629-1635. <https://doi.org/10.1007/s40279-019-01134-5>
- Cesar, G. M., Havens, K. L., Chang, Y.-J., & Sigward, S. M. (2011). Effects of cut angle and online processing on cutting maneuvers. (Ed.),^(Eds.). American Society of Biomechanics Annual Conference.
- Decker, M. J., Torry, M. R., Wyland, D. J., Sterett, W. I., & Richard Steadman, J. (2003). Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. *Clin Biomech (Bristol, Avon)*, 18(7), 662-669. [https://doi.org/10.1016/s0268-0033\(03\)00090-1](https://doi.org/10.1016/s0268-0033(03)00090-1)
- Draganich, L. F., Andriacchi, T. P., & Andersson, G. B. (1987). Interaction between intrinsic knee mechanics and the knee extensor mechanism. *Journal of Orthopaedic Research*, 5(4), 539-547. <https://doi.org/10.1002/jor.1100050409>
- The female ACL: Why is it more prone to injury? (2016). *J Orthop*, 13(2), A1-A4. [https://doi.org/10.1016/S0972-978X\(16\)00023-4](https://doi.org/10.1016/S0972-978X(16)00023-4)
- Ford, K. R., Myer, G. D., & Hewett, T. E. (2003). Valgus knee motion during landing in high school female and male basketball players. *Medicine and Science in Sports and Exercise*, 35(10), 1745-1750. <https://doi.org/10.1249/01.Mss.0000089346.85744.D9>
- Gandevia, S. C. (2001). Spinal and supraspinal factors in human muscle fatigue. *Physiol Rev*, 81(4), 1725-1789.
- Gribble, P. A., Hertel, J., Denegar, C. R., & Buckley, W. E. (2004). The Effects of Fatigue and Chronic Ankle Instability on Dynamic Postural Control. *J Athl Train*, 39(4), 321-329.
- Harmon, K. G., & Ireland, M. L. (2000). Gender differences in noncontact anterior cruciate ligament injuries. *Clinics in sports medicine*, 19(2), 287-302.
- Herrington, L. (2011). Knee valgus angle during landing tasks in female volleyball and basketball players. *The Journal of Strength & Conditioning Research*, 25(1), 262-266.
- Hewett, T. E., Ford, K. R., & Myer, G. D. (2005). Anterior cruciate ligament injuries in female athletes - Part 2, a meta-analysis of neuromuscular interventions aimed at injury prevention. *American Journal of Sports Medicine*, 34(3), 490-498. <https://doi.org/10.1177/0363546505282619>
- Hewett, T. E., & Myer, G. D. (2011). The mechanistic connection between the trunk, hip, knee, and anterior cruciate ligament injury. *Exerc Sport Sci Rev*, 39(4), 161-166. <https://doi.org/10.1097/JES.0b013e3182297439>
- Hewett, T. E., Torg, J. S., & Boden, B. P. (2009). Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: lateral trunk and knee abduction motion are combined components of the injury mechanism. *British Journal of Sports Medicine*, 43(6), 417-422.

- Hiemstra, L. A., Lo, I. K., & Fowler, P. J. (2001). Effect of fatigue on knee proprioception: implications for dynamic stabilization. *J Orthop Sports Phys Ther*, 31(10), 598-605. <https://doi.org/10.2519/jospt.2001.31.10.598>
- Koga, H., Nakamae, A., Shima, Y., Iwasa, J., Myklebust, G., Engebretsen, L., Bahr, R., & Krosshaug, T. (2010). Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. *Am J Sports Med*, 38(11), 2218-2225. <https://doi.org/10.1177/0363546510373570>
- Kristianslund, E., Faul, O., Bahr, R., Myklebust, G., & Krosshaug, T. (2014). Sidestep cutting technique and knee abduction loading: implications for ACL prevention exercises. *British Journal of Sports Medicine*, 48(9), 779-783. <https://doi.org/DOI10.1136/bjsports-2012-091370>
- Lattanzio, P. J., Petrella, R. J., Sproule, J. R., & Fowler, P. J. (1997). Effects of fatigue on knee proprioception. *Clin J Sport Med*, 7(1), 22-27. <https://doi.org/10.1097/00042752-199701000-00005>
- Laughlin, W. A., Weinhandl, J. T., Kernozek, T. W., Cobb, S. C., Keenan, K. G., & O'Connor, K. M. (2011). The effects of single-leg landing technique on ACL loading. *Journal of Biomechanics*, 44(10), 1845-1851. <https://doi.org/10.1016/j.jbiomech.2011.04.010>
- Liederbach, M., Kremenec, I. J., Orishimo, K. F., Pappas, E., & Hagins, M. (2014). Comparison of landing biomechanics between male and female dancers and athletes, part 2: Influence of fatigue and implications for anterior cruciate ligament injury. *Am J Sports Med*, 42(5), 1089-1095. <https://doi.org/10.1177/0363546514524525>
- Makaruk, H., Czaplicki, A., Sacewicz, T., & Sadowski, J. (2014). The effects of single versus repeated plyometrics on landing biomechanics and jumping performance in men. *Biol Sport*, 31(1), 9-14. <https://doi.org/10.5604/20831862.1083273>
- Mantovani, G., & Lamontagne, M. (2017). How Different Marker Sets Affect Joint Angles in Inverse Kinematics Framework. *J Biomech Eng*, 139(4), 1-7. <https://doi.org/10.1115/1.4034708>
- McLean, S. G., Fellin, R. E., Suedekum, N., Calabrese, G., Passerallo, A., & Joy, S. (2007). Impact of fatigue on gender-based high-risk landing strategies. *Med Sci Sports Exerc*, 39(3), 502-514. <https://doi.org/10.1249/mss.0b013e3180d47f0>
- McLean, S. G., Huang, X., & van den Bogert, A. J. (2005). Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: implications for ACL injury. *Clin Biomech (Bristol, Avon)*, 20(8), 863-870. <https://doi.org/10.1016/j.clinbiomech.2005.05.007>
- McLean, S. G., Lipfert, S. W., & van den Bogert, A. J. (2004). Effect of gender and defensive opponent on the biomechanics of sidestep cutting. *Med Sci Sports Exerc*, 36(6), 1008-1016. <https://doi.org/10.1249/01.mss.0000128180.51443.83>
- Miura, K., Ishibashi, Y., Tsuda, E., Okamura, Y., Otsuka, H., & Toh, S. (2004). The effect of local and general fatigue on knee proprioception. *Arthroscopy*, 20(4), 414-418. <https://doi.org/10.1016/j.arthro.2004.01.007>

- Norcross, M. F., Blackburn, J. T., Goerger, B. M., & Padua, D. A. (2010). The association between lower extremity energy absorption and biomechanical factors related to anterior cruciate ligament injury. *Clinical Biomechanics*, 25(10), 1031-1036. <https://doi.org/10.1016/j.clinbiomech.2010.07.013>
- Nunley, R. M., Wright, D., Renner, J. B., Yu, B., & Garrett Jr, W. E. (2003). Gender comparison of patellar tendon tibial shaft angle with weight bearing. *Research in Sports Medicine*, 11(3), 173-185.
- Nyland, J. A., Caborn, D. N., Shapiro, R., & Johnson, D. L. (1997). Fatigue after eccentric quadriceps femoris work produces earlier gastrocnemius and delayed quadriceps femoris activation during crossover cutting among normal athletic women. *Knee Surg Sports Traumatol Arthrosc*, 5(3), 162-167. <https://doi.org/10.1007/s001670050045>
- O'Connor, K. M., Johnson, C., & Benson, L. C. (2015). The Effect of Isolated Hamstrings Fatigue on Landing and Cutting Mechanics. *J Appl Biomech*, 31(4), 211-220. <https://doi.org/10.1123/jab.2014-0098>
- Olsen, O. E., Myklebust, G., Engebretsen, L., & Bahr, R. (2004). Injury mechanisms for anterior cruciate ligament injuries in team handball a systematic video analysis. *American Journal of Sports Medicine*, 32(4), 1002-1012. <https://doi.org/10.1177/0363546503261724>
- Olsen, O. E., Myklebust, G., Engebretsen, L., & Bahr, R. (2006). Injury pattern in youth team handball: a comparison of two prospective registration methods. *Scand J Med Sci Sports*, 16(6), 426-432. [https://doi.org/DOI 10.1111/j.1600-0838.2005.00484.x](https://doi.org/DOI%2010.1111/j.1600-0838.2005.00484.x)
- Orishimo, K. F., & Kremenec, I. J. (2006). Effect of fatigue on single-leg hop landing biomechanics. *J Appl Biomech*, 22(4), 245-254. <https://doi.org/10.1123/jab.22.4.245>
- Pataky, T. C. (2010). Generalized n-dimensional biomechanical field analysis using statistical parametric mapping. *Journal of Biomechanics*, 43(10), 1976-1982. <https://doi.org/10.1016/j.jbiomech.2010.03.008>
- Sigward, S. M., Cesar, G. M., & Havens, K. L. (2015). Predictors of Frontal Plane Knee Moments During Side-Step Cutting to 45 and 110 Degrees in Men and Women: Implications for Anterior Cruciate Ligament Injury. *Clin J Sport Med*, 25(6), 529-534. <https://doi.org/10.1097/jsm.0000000000000155>
- Sigward, S. M., & Powers, C. M. (2007). Loading characteristics of females exhibiting excessive valgus moments during cutting. *Clin Biomech (Bristol, Avon)*, 22(7), 827-833. <https://doi.org/10.1016/j.clinbiomech.2007.04.003>
- Thorlund, J. B., Michalsik, L. B., Madsen, K., & Aagaard, P. (2008). Acute fatigue-induced changes in muscle mechanical properties and neuromuscular activity in elite handball players following a handball match. *Scand J Med Sci Sports*, 18(4), 462-472. <https://doi.org/10.1111/j.1600-0838.2007.00710.x>
- Wang, L. I. (2011). The lower extremity biomechanics of single- and double-leg stop-jump tasks. *J Sports Sci Med*, 10(1), 151-156.

- Wojtys, E. M., Huston, L. J., Schock, H. J., Boylan, J. P., & Ashton-Miller, J. A. (2003). Gender differences in muscular protection of the knee in torsion in size-matched athletes. *J Bone Joint Surg Am*, 85(5), 782-789. <https://doi.org/10.2106/00004623-200305000-00002>
- Wojtys, E. M., Wylie, B. B., & Huston, L. J. (1996). The effects of muscle fatigue on neuromuscular function and anterior tibial translation in healthy knees. *Am J Sports Med*, 24(5), 615-621. <https://doi.org/10.1177/036354659602400509>
- Xia, R., Zhang, X., Wang, X., Sun, X., & Fu, W. (2017). Effects of Two Fatigue Protocols on Impact Forces and Lower Extremity Kinematics during Drop Landings: Implications for Noncontact Anterior Cruciate Ligament Injury. *J Healthc Eng*, 2017, 5690519-5690519. <https://doi.org/10.1155/2017/5690519>
- Yu, B., & Garrett, W. E. (2007). Mechanisms of non-contact ACL injuries. *Br J Sports Med*, 41 Suppl 1, i47-51. <https://doi.org/10.1136/bjsm.2007.037192>
- Zech, A., Steib, S., Hentschke, C., Eckhardt, H., & Pfeifer, K. (2012). Effects of localized and general fatigue on static and dynamic postural control in male team handball athletes. *J Strength Cond Res*, 26(4), 1162-1168. <https://doi.org/10.1519/JSC.0b013e31822dfbbb>

CAPÍTULO 4

ESTUDO 3

A custom musculoskeletal model for estimation of medial and lateral tibiofemoral contact forces during tasks with high knee and hip flexions

Bruno Luiz Souza Bedo; Danilo S. Catelli; Mario Lamontagne; Paulo Roberto Pereira Santiago

A custom musculoskeletal model for estimation of medial and lateral tibiofemoral contact forces during tasks with high knee and hip flexions

Bruno Luiz Souza Bedo^{A,B,C}; Danilo S. Catelli^B; Mario Lamontagne^B; Paulo Roberto Pereira Santiago^{A,C}

^ARibeirao Preto Medical School, University of Sao Paulo, Ribeirao Preto, SP, Brazil;

^BSchool of Human Kinetics, University of Ottawa, Ottawa, ON, Canada;

^CSchool of Physical Education and Sports of Ribeirão Preto, University of São Paulo, Ribeirão Preto, SP, Brazil

A versão publicada pode ser acessada clicando [aqui](#) ou pelo QR code abaixo.¹³

Favor citar:

Bedo, B. L. S., Catelli, D. S., Lamontagne, M., & Santiago, P. R. P. (2020). A custom musculoskeletal model for estimation of medial and lateral tibiofemoral contact forces during tasks with high knee and hip flexions. *Computer Methods in Biomechanics and Biomedical Engineering*, 1-6.

¹³ Em caso de dificuldade para acessar o arquivo, segure a tecla “ctrl” e clique no link abaixo.

Link: <https://www.tandfonline.com/doi/full/10.1080/10255842.2020.1757662>

Se o problema persistir, por favor envie um e-mail para: bruno.bedo@usp.br ou paulosantiag@usp.br.

Abstract

Most of musculoskeletal models (MSKM) estimate the tibiofemoral joint reaction load at a single point or do not support large lower-limb ranges. This study aimed to adapt a generic MSKM that allows large knee and hip flexions to compute medial and lateral tibiofemoral contact forces (TFCF) during gait and squat tasks. The updated model includes medial and lateral knee compartment geometries that allow computing the vertical TFCF. The updated MSKM does not affect kinematics and kinetics outputs in both of the tasks, and the sum of the medial and lateral TFCF was equivalent to the net TFCF of the original MSKM.

Keywords: Musculoskeletal model; tibiofemoral; knee; knee contact forces; squat; gait.

Introduction

The inverse dynamic approach has been used to indirectly measure the load on the medial and lateral condyles (D. G. Lloyd & Besier, 2003); however, this approach still does not take account for muscle forces. Tibiofemoral alignment asymmetry affects the load redistribution on the medial and lateral components, increasing the risks of osteoarthritis (Jackson, Wluka, Teichtahl, Morris, & Cicuttini, 2004). Therefore, estimation of knee contact loading in the medial and lateral compartments would provide a beneficial assessment (Manal & Buchanan, 2013).

Musculoskeletal models (MSKM) have been efficient to investigate human movement and predict the effect of interventions in a non-invasive fashion (Delp et al., 2007) as well as to estimate muscle forces and joint contact loading (Catelli, Wesseling, Jonkers, & Lamontagne, 2019b; Lerner et al., 2015; Rajagopal et al., 2016). In order to account for a better muscle force solution in a young cohort, Rajagopal et al. (2016) used magnetic resonance imaging of young healthy individuals to redefine musculotendon parameters (Figure 1b). This MSKM model was further revised twice, when increasing the knee range and updating musculotendon units (MTU) force-generating properties and paths (Figure 1c) (Lai et al. (2017), and by controlling the moment arm of hip and knee muscles to enable investigations with high range of motion (Figure 1d) (Catelli, Wesseling, et al., 2019b). Still, these MSKMs calculate the knee contact load at a single point in the tibiofemoral joint. Resolving the magnitudes of medial and lateral forces by approximating compartment points of contact is an approach for estimating contact forces (Gerus et al., 2013; Kumar, Rudolph, & Manal, 2012; Winby, Lloyd, Besier, & Kirk, 2009), which has been already implemented (Lerner et al. 2015) in a preceding MSKM (Delp et al. 1990), that still lacks muscle paths reliability during high lower-limb flexions and has its musculotendon parameters based on cadaveric data (Figure 1a).

Figure 1. The schematic musculoskeletal models (MSKM) that have been used in lower limb research and the implementation in the updated model. (A) Allows quantifying medial and lateral tibiofemoral contact forces. (B) Implemented musculotendon unit (MTU) parameters based on healthy and young individuals. (C) Increased knee range of motion and updated knee MTU parameters and path. (D) Controlled hip MTU moment arms to allow higher hip and knee flexion. The updated model (E) implemented the medial and lateral tibiofemoral contact forces strategy in an MSKM based in young and healthy MTU parameters and that allows a high range of motion of the hip and knee.

Therefore, the purpose of this study was to examine the effect of including the method that computes medial and lateral TFCF to an MSKM that allows large lower-limb joint ranges and test its reliability during gait and deep squatting tasks. We hypothesized that the sum of the medial and lateral vertical components of the TFCF of the updated MSKM should match with the single vertical TFCF of the original MSKM during both tasks, as well as the joint kinematics and kinetics outcomes should not differ between the two models.

Methods

Generic model

The generic MSKM used in this study has 80 lower-limb Hill-type MTU, 37 degrees of freedom, and 17 torque actuators driving the upper body (Rajagopal et al., 2016). The MTU parameters were based on the combination of MRI cadaver and in-vivo data (Rajagopal et al., 2016), had its knee paths and some force-generating properties updated (Lai et al., 2017), and had some addition wrapping surface implemented at the knee and hip joints in order to control the MTU moment arms during deep squatting task (Catelli, Wesseling, et al., 2019b). The analyses were all processed in OpenSimTM 3.3 (Stanford University, Stanford, USA) (Delp et al., 2007).

Subsequently, knee geometry adaptations were implemented to allow the quantification of medial and lateral TFCF, according to a previous generic MSKM Lerner et al. (2015). This MSKM used tibial and femoral components to resolve the differences between medial and lateral TFCF, configured the frontal-plane alignment of the knee joint and also shared all loads transmitted between the femur and tibia, resolving them as the medial and lateral contact forces (Lerner et al., 2015).

Modification to the model

The updated MSKM knee joint included a femoral component acting directly to the sagittal articulation frame, with two revolute joints connecting the parent (*sagittal articulation frame*) and to the child (*medial and lateral compartments*). In addition, one weld joint (d_1 - *medial*) and one weld constraint (d_2 - *lateral*) connects the medial and lateral components to the tibial plateau that, in turn, is connected to the tibia through another weld joint (θ_2) - Figure 1a. (Supplement Figure 1). The two revolute joints, connecting the sagittal articulation frame to the medial and lateral tibiofemoral compartments, share the load transmitted between the femur and tibia and resolve them as medial and lateral contact forces (Lerner et al., 2015).

Motion capture

Gait and deep squat modelling were performed by tracking experimental data from a healthy male participant (35.0 years, 78.2 kg, and 176 cm). The three-dimensional motion capture system included 10 infrared cameras at 200 Hz (MX-13, Vicon, Oxford, UK), that tracked 45 full-body marker trajectories (Mantovani & Lamontagne, 2017b), and two force plates (1000 Hz, Bertec, Columbus, USA).

For the gait task, the analyses were performed during the stance phase (i.e., foot-strike to foot-off). For the deep squat task, the participant stood with each foot on a force plate with feet hip-width apart and was instructed to squat as low as possible. The squat cycle was defined from standing to the deepest squat and back to the standing position. The markers trajectories were labelled and along with the ground reaction forces, were filtered in Nexus 2.5 (Vicon, Oxford, UK) and imported into OpenSim.

Data analysis and model evaluation

The simulations of both tasks were performed using the original (Catelli, Wesseling, et al., 2019b) and updated MSKM. Both MSKM were scaled based on a static pose, the joint angles and the net joint moments were computed using the inverse kinematics and the inverse dynamics tools, respectively. Muscle activations were calculated using static optimization while minimizing the sum of squared muscle activations. The joint reaction forces were computed using resultant forces and moments acting on each articulating joint from all muscle forces and also the internal and external load applied to the model. The medial and lateral TFCF were calculated using OpenSim's *JointReaction* analyses. The knee contact impulse was calculated based on the integral of the vertical TFCF in both MSKM in order to compare the net TFCF of the two models.

Results

Minimal differences in kinematics (max difference: 1.8° in the sagittal plane for the ankle joint) and kinetics (max difference: 0.42 N.mm in the frontal plane for the hip joint) were found between the two models during the gait or the squat tasks (Supplement Table 1). Muscle forces of both models were also similar, with the highest difference between models found for the muscle *semimembranosus* (0.04 BW) and *rectus femoris* (0.02 BW) for gait and squat, respectively (Supplement Table 2).

The two revolute joints inserted on medial and lateral condyles actuated on the frontal plane connecting the medial and lateral components to the sagittal articulation frame enabled the vertical medial and lateral TFCF quantification. The sum of the medial and lateral vertical components of the updated MSKM nearly matched the vertical component of the original MSKM, with the highest difference of 0.07 BW for gait and 0.04 BW for squat.

During the gait task, the medial condyle showed a larger contact force (peak = 2.60 BW) in comparison to lateral one (peak = 0.69 BW; Figure 2a). This difference can also be inferred on the compass graph generated from the original MSKM, where its peak points towards the medial direction, corroborating with the descriptive medial and lateral results of the updated MSKM (Figure 2e).

During the deep squat, a larger force was found in the lateral knee component (peak = 4.61 BW) in comparison with the medial (peak = 2.40 BW), which also corroborates with the compass plotting, where the peak contact force is directed to the lateral compartment of the knee (Figure 2b and 2d). In addition, the integral of the vertical TFCF curve of the original MSKM matched with the sum of the areas of the medial and lateral components from the updated MSKM in both gait and squat tasks (Figure 2e and 2f).

Figure 2. Tibiofemoral contact forces (TFCF) during the gait (top) and squat (bottom) cycles for the original (blue) and updated (red) musculoskeletal model. The graphs on the left column represent the medial (top) and lateral (bottom) components reaction forces during the gait (A) and squat (B) cycles for the updated model. The graphs in the middle show the joint contact impulse calculated by the integral of TFCF from the original and updated models during the gait (C) and squat (D). The dark and light red columns represent the medial and lateral components, respectively. Lastly, the graphs on the right column show the body weight (BW) normalized TFCF resultant during the gait (E) squat (F) cycle, reported with the compass graph (posterior view). The blue lines show the magnitude (radar graph) and direction to the tibial plateau obtained from the original model. The red arrows represent the inverse of the peak TFCF of the lateral (light red) and medial (dark red) components obtained from the updated model.

Discussion

The purpose of this study was to update an MSKM that was recently customized to allow high knee and hip flexions (Catelli, Wesseling, et al., 2019b) to quantify the TFCF in the medial and lateral condyles. The results demonstrated that the updated MSKM successfully quantified the medial and lateral force components for gait and squat tasks without compromising kinematic, kinetic or muscle force analysis. Equally to the original MSKM, the updated MSKM also allows maximum knee and hip flexions of 145° and 138°, respectively.

The updated model showed higher TFCF in the medial component during gait and higher TFCF in the lateral component during squat, which could be inferred by the compass plots showing the TFCF directions. These results provide an important contribution toward understanding the joint load on the knee joint during gait and squat tasks, that can also be helpful in clinical cases (e.g. knee osteoarthritis patients).

The similarity between the kinematics, kinetics and muscle forces confirms that the muscles and joints geometries were not affected by the changes performed on the updated MSKM. The sum of both medial/lateral condyles vertical load matched with the vertical component quantified on the original MSKM, as well as the sum of the medial/lateral joint contact impulse matched the single joint contact impulse of the original MSKM. The larger medial contact force found during gait is likely a consequence of a medial deviation on the transversal axis during gait (Zhao, Banks, D'Lima, Colwell, & Fregly, 2007). Differently, the larger lateral contact force during squat can be associated with the tibiofemoral orientation during the extreme knee flexion occurring during the task (i.e. deep squat), as the orientation between the femur (*parent*) and the tibia (*child*) on the original MSKM undergoes an internal rotation while the knee flexion increases. Therefore, it is plausible that the larger knee flexion increases the forces in the medial contact, once the tibiofemoral alignment affects the knee joint load distribution (Lerner et al., 2015).

Model development and validation experiments are closely linked to (Hicks et al., 2015). The original model for this study (Rajagopal et al., 2016) presented the validation with the inclusion of successful simulations of walking and running, that showed a qualitative agreement of the simulated muscle activity with experimental electromyography data. Considering that joint contact forces estimation is sensitive to both the joint kinematic definition and the distribution of muscle forces crossing the joint in that motion (DeMers et al., 2014; Hicks et al., 2015), the comparability between these models assures that the joint contact forces of the updated one are reliable. Future work is still necessary to validate both of these components.

This updated model brings an important contribution toward understanding the medial and lateral knee loading compartments during movements involving high hip and knee flexions. Still, individual examination after scaling the model is strongly recommended, since the distance between the condyle is scaling dependent. This model is available from SimTK.org (<https://simtk.org/projects/knee-contact>), and we encourage others to make refinements and share them with the researchers of the computational modeling area.

Disclosure statement

No potential conflict of interest was reported by the authors.

Funding

This study was partially financed by the Coordenação de Aperfeiçoamento de Pessoal de Nível Superior – Brasil (CAPES) – Finance Code 001; São Paulo Research Foundation – FAPESP, Brazil [2019/04721-0]; Canadian Institutes of Health Research (CIHR) - grant number: PJT-162333; and the International Society of Biomechanics: Developing Countries Grant Competition 2019.

References

- Catelli DS, Wesseling M, Jonkers I, Lamontagne M. 2019. A musculoskeletal model customized for squatting task. *Computer methods in biomechanics and biomedical engineering*. 22(1):21-24. eng.
- Delp SL, Anderson FC, Arnold AS, Loan P, Habib A, John CT, Guendelman E, Thelen DG. 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE transactions on bio-medical engineering*. 54(11):1940-1950. eng.
- Gerus P, Sartori M, Besier TF, Fregly BJ, Delp SL, Banks SA, Pandy MG, D'Lima DD, Lloyd DG. 2013. Subject-specific knee joint geometry improves predictions of medial tibiofemoral contact forces. *J Biomech*. 46(16):2778-2786. eng.
- Hicks JL, Uchida TK, Seth A, Rajagopal A, Delp SL. 2015. Is my model good enough? Best practices for verification and validation of musculoskeletal models and simulations of movement. *J Biomech Eng*. 137(2):020905. eng.
- Jackson BD, Wluka AE, Teichtahl AJ, Morris ME, Cicuttini FM. 2004. Reviewing knee osteoarthritis--a biomechanical perspective. *J Sci Med Sport*. 7(3):347-357. eng.
- Kumar D, Rudolph KS, Manal KT. 2012. EMG-driven modeling approach to muscle force and joint load estimations: case study in knee osteoarthritis. *J Orthop Res*. 30(3):377-383. eng.
- Lai AKM, Arnold AS, Wakeling JM. 2017. Why are Antagonist Muscles Co-activated in My Simulation? A Musculoskeletal Model for Analysing Human Locomotor Tasks. *Annals of biomedical engineering*. 45(12):2762-2774. eng.
- Lerner ZF, DeMers MS, Delp SL, Browning RC. 2015. How tibiofemoral alignment and contact locations affect predictions of medial and lateral tibiofemoral contact forces. *J Biomech*. 48(4):644-650. eng.
- Lloyd DG, Besier TF. 2003. An EMG-driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo. *J Biomech*. 36(6):765-776. eng.
- Manal K, Buchanan TS. 2013. An electromyogram-driven musculoskeletal model of the knee to predict in vivo joint contact forces during normal and novel gait patterns. *J Biomech Eng*. 135(2):021014. eng.
- Mantovani G, Lamontagne M. 2017. How Different Marker Sets Affect Joint Angles in Inverse Kinematics Framework. *J Biomech Eng*. 139(4). eng.
- Rajagopal A, Dembia CL, DeMers MS, Delp DD, Hicks JL, Delp SL. 2016. Full-Body Musculoskeletal Model for Muscle-Driven Simulation of Human Gait. *IEEE transactions on bio-medical engineering*. 63(10):2068-2079. eng.
- Winby CR, Lloyd DG, Besier TF, Kirk TB. 2009. Muscle and external load contribution to knee joint contact loads during normal gait. *J Biomech*. 42(14):2294-2300. eng.
- Zhao D, Banks SA, D'Lima DD, Colwell CW, Jr., Fregly BJ. 2007. In vivo medial and lateral tibial loads during dynamic and high flexion activities. *J Orthop Res*. 25(5):593-602. eng.

Supplement Table 1. Joint range of motion and moment between the original and updated MSKM during gait and squat tasks. The results consider the left side as the side of interest.

Joint kinematics					
Joint	Plane	Gait (deg)		Squat (deg)	
		Original	Updated	Original	Updated
Hip	Sagittal	51.24	51.25	116.58	116.68
	Frontal	14.02	13.93	10.44	10.44
	Transversal	10.16	10.09	21.10	21.13
Knee	Sagittal	24.05	24.09	117.60	117.66
Ankle	Sagittal	29.46	29.54	27.96	28.05
Joint kinetics					
Joint	Plane	Gait (N.mm)		Squat (N.mm)	
		Original	Updated	Original	Updated
Hip	Sagittal	86.26	86.26	74.29	74.20
	Frontal	84.84	84.96	16.09	16.07
	Transversal	18.94	18.99	19.80	19.81
Knee	Sagittal	75.36	75.56	72.31	72.30
Ankle	Sagittal	107.84	107.92	36.08	26.09

Supplement Table 2. Peak muscle forces and the maximum difference (MD) during gait and deep squat tasks. The results consider the left side as the side of interest.

Muscle forces (BW)						
Muscles	Gait			Squat		
	Peak			Peak		
	Original	Updated	MD	Original	Updated	MD
gasmed	1.625	1.624	0.009	0.464	0.463	0.008
gaslat	0.586	0.587	0.003	0.184	0.183	0.003
glmed1	0.965	0.970	0.035	0.061	0.061	0.001
glmed2	0.192	0.193	0.002	0.042	0.417	0.002
glmed3	0.102	0.103	0.000	0.122	0.121	0.002
glmax1	0.076	0.076	0.002	0.148	0.148	0.002
glmax2	0.175	0.174	0.003	0.824	0.829	0.006
glmax3	0.186	0.186	0.000	0.421	0.422	0.003
iliacus	1.085	1.087	0.040	0.030	0.031	0.003
perlong	0.605	0.613	0.007	0.073	0.073	0.002
psoas	1.133	1.136	0.031	0.072	0.073	0.003
bflhl	0.173	0.174	0.011	0.845	0.844	0.007
soleus	1.653	1.655	0.016	0.777	0.778	0.008
recfem	0.649	0.651	0.005	0.300	0.322	0.023
semimem	0.497	0.539	0.046	1.523	1.523	0.001
vaslat	0.941	0.942	0.033	4.938	4.931	0.017
vasmed	0.025	0.253	0.004	1.408	1.409	0.005
tibant	0.334	0.334	0.008	0.231	0.230	0.001

Note: *gasmed*: gastrocnemius medialis; *gaslat*: gastrocnemius lateralis; *glmed1*, *glmed2* and *glmed3*: gluteus medialis; *glmax1*, *glmax2* and *glmax3*: gluteus maximum; *perlong*: peroneus longus; *bflhl*: biceps femoris long head; *recfem*: rectus femoris; *semimem*: semimembranosus; *vaslat*: vastus laealis; *vasmed*: vastus medialis; *tibant*: tibialis anterior. The MD was calculated by the difference between the full-wave muscle force of the original and updated model.

Supplement Figure 1. The topology view exported from OpenSim3.3® of the updated model. The right and left knee joints are composed by the medial (*med_cond_joint_r* and *med_cond_joint_l*) and lateral (*lat_cond_joint_r* and *med_cond_joint_l*) joints that share all tibiofemoral load on both components. The sagittal knee angles are based on the knee joint (*knee_r* and *knee_l*) that are connected between the parent (*femoral_cond_r* and *femoral_cond_l*) and the child (*sagittal_articulation_frame_r* and *sagittal_articulation_frame_l*).

CAPÍTULO 5

ESTUDO 4

BOPS: A Matlab toolbox to Batch Musculoskeletal Data Processing from OpenSim

Bruno L. S. Bedo; Alice Mantoan; Danilo S. Catelli; Willian Cruaud; Monica Reggiani; Mario Lamontagne

BOPS: A Matlab toolbox to Batch Musculoskeletal Data Processing from OpenSim

Bruno L. S. Bedo^{A,B}; Alice Mantoan^C; Danilo S. Catelli^B; Willian Cruaud^B; Monica Reggiani^C; Mario Lamontagne^B

^ARibeirao Preto Medical School, University of Sao Paulo, Ribeirao Preto, Brazil.

^BFaculty of Health Sciences, University of Ottawa, Ottawa, Canada.

^CDepartment of Management and Engineering, University of Padova, Vicenza, Italy.

* *B.L.S.B., A.M. and D.S.C. contributed equally as first authors.*

A versão publicada pode ser acessada clicando [aqui](#) ou pelo QR code abaixo.¹⁴

Favor citar:

Bruno L. S. Bedo, Alice Mantoan, Danilo S. Catelli, Willian Cruaud, Monica Reggiani & Mario Lamontagne (2021) BOPS: a Matlab toolbox to batch musculoskeletal data processing for OpenSim, Computer Methods in Biomechanics and Biomedical Engineering, DOI: 10.1080/10255842.2020.1867978

¹⁴ Em caso de dificuldade para acessar o arquivo, segure a tecla “ctrl” e clique no link abaixo.

Link: <https://www.tandfonline.com/doi/full/10.1080/10255842.2020.1867978>

Se o problema persistir, por favor envie um e-mail para: bruno.bedo@usp.br ou paulosantiag@usp.br

Abstract

This paper presents Batch OpenSim Processing Scripts (BOPS), a Matlab toolbox for batch processing common OpenSim procedures: Inverse Kinematics, Inverse Dynamics, Muscle Analysis, Static Optimization, and Joint Reaction Analysis. BOPS is an easy-to-use and highly configurable tool that aims to reduce the time required to process large datasets, thus fostering the adoption of musculoskeletal modeling and simulations in daily practice. Its graphical user interface includes pre-defined setup files and has been designed to fulfill the needs of different research projects by simplifying the the customization of the procedures, facilitating the analysis, and boosting research group collaborations. BOPS is released under Apache License 2.0, and its source code is freely available on SimTK and GitHub.

Keywords: Musculoskeletal simulations, OpenSim, Data Processing, Matlab toolbox.

Introduction

Musculoskeletal modeling provides a non-invasive approach to estimate muscle forces and joint contact loading, and it is an essential tool to better understand the musculoskeletal system's actions during human movement (Delp et al. 2007; Seth et al. 2018). However, it usually requires to collect motion data and to process and export them to obtain input data for the simulations. Besides, applications may require personalized musculoskeletal models, which thus must be generated. Commercial solutions and free tools are available to the biomechanics community to simplify each step within the workflow of neuro- and musculoskeletal modeling (Figure 1). For instance, BTK (Barre & Armand 2014) allows the reading and writing of C3D files, the standard file format in the field (C3D – Coordinate 3D, Motion Lab Systems), while Mokka, a standalone application by the same authors, was conceived for their visual inspection; the MoCap (Burger & Toiviainen 2013) and biomechZoo (Dixon et al. 2017) are two toolboxes enabling motion data analysis and visualization; ADAT toolbox (James & Wixted 2011) is intended to handle sports movement data; pyomeca (Martinez et al. 2020) enables data extraction, processing and visualization for use in research and education, and MOtoNMS (Mantoan et al. 2015) processes experimental data from different motion capture devices and generates inputs for neuromusculoskeletal modeling. CEINMS is the first open-source toolbox that allows individual neural drive from EMG to be used as input into simulations (Pizzolato et al. 2015), while NMS Builder was conceived as a free software application to generate subject-specific musculoskeletal models for OpenSim (Valente et al. 2017). The open-source Musculoskeletal Atlas Project (MAP) Client software provides workflows for generating musculoskeletal models from clinical data using statistical models (Zhang et al. 2014).

Figure 1. Workflow for neuro- and musculoskeletal modeling applications and list of toolboxes available to accomplish each step. Open-source toolboxes are highlighted in bold.

Most of the listed tools have been developed in Matlab (Mathworks Inc., Natick, USA), indeed only those requiring high computational performances (CEINMS) or conceived as a standalone application (i.e., NMS Builder and OpenSim) are based on C++ programming language (Table 1). Therefore, Matlab turns out to be the preferred development tool for data processing and analysis in biomechanics, likely due to its general availability within research groups and the basic programming skills required.

Although various commercial software tools to analyze human motion through computational modeling and simulation of musculoskeletal systems are available (Davoodi et al. 2003; Damsgaard et al. 2006, Figure 1 - red box), researchers often choose OpenSim (Stanford University, USA) for its open-source availability and the large community of users (Delp et al. 2007; Seth et al. 2018). OpenSim enables the estimation of joint angles, moments, muscle and joint contact forces for different movement tasks (Lerner et al. 2015; Rajagopal et al. 2016; Catelli, Wesseling, et al. 2019; Bedo et al. 2020). However, it often requires an extensive number of steps to process even a single trial¹⁵, which increases the risk of blunders when considering multiple trials, and also results in long processing time.

Although Mansouri & Reinbolt (2012) proposed a Matlab/Simulink tool based on OpenSim to perform forward dynamics simulation of OpenSim models with feedback control, to the authors' knowledge, there is yet no open-access tool available to time-optimize data processing and outcome analysis in OpenSim.

Therefore, this paper presents Batch OpenSim Processing Scripts¹⁶ (BOPS), an easy-to-use Matlab toolbox provided with a graphical user interface (GUI) to batch process complex biomechanics dataset with the following OpenSim procedures: *Inverse Kinematics*, *Inverse Dynamics*, *Muscle Analysis*, *Static Optimization*, and *Joint Reaction Analysis*.

¹⁵ OpenSim User's Guide: simtk-confluence.stanford.edu:8443/display/OpenSim33/User%27s+Guide

¹⁶ Batch OpenSim Processing Scripts: github.com/RehabEngGroup/BOPS or simtk.org/projects/bops

Table 1. Details on open-source toolboxes in Figure 1 (colors on the left column refer to the corresponding step and toolbox main goal depicted in Figure 1). The used platform, the availability of Test Data, and the License are compared. Links to the project pages, software repositories, and available documentation are included.

Toolbox	Platform	Project Page	Repository	Documentation	Test Data	License
biomechZoo	Matlab	biomechzoo.com/	github.com/PhilD001/biomechZoo	x	x	Apache v2.0
BTK, Mokka	C++	code.google.com/archive/p/btk/	github.com/Biomechanical-ToolKit	biomechanical-toolkit.github.io/	V	New BSD license
MoCap	Matlab	jyu.fi/music/coe/materials/mocaptoolbox	X	x	x	GPL
ADAT	Matlab	adat.gsportstechnology.com/	X	x	x	Creative Commons
pyomeca	Python	github.com/pyomeca/	github.com/pyomeca/pyomeca/	github.com/pyomeca/Documentation	V	Creative Commons
MOtoNMS	Matlab	simtk.org/home/motonms/	github.com/RehabEngGroup/MOtoNMS	rehabenggroup.github.io/MOtoNMS/	V	GNU General Public License v3
NMS Builder	C++	nmsbuilder.org/	X	www.nmsbuilder.org/downloads/nms_builder_manual_102.pdf	V	ALBA and OpenSim Library Licence
MAP Client	Python	map-client.readthedocs.io/en/latest/	github.com/MusculoskeletalAtlasProject/mapclient	map-client.readthedocs.io/en/latest/developer/index.html	V	GNU General Public License v3
OpenSim	C++	simtk.org/projects/opensim	github.com/opensim-org	simtk-confluence.stanford.edu/display/OpenSim/OpenSim+Documentation	V	Apache v 2.0
CusToM	Matlab	github.com/anmuller/CusToM	github.com/anmuller/CusToM	github.com/anmuller/CusToM/blob/master/Docs/CusToMDocumentation.pdf	V	GPL-3.0
CEINMS	C++	simtk.org/projects/ceinms	github.com/CEINMS/CEINMS	simtk.org/frs/?group_id=840	V	Apache v 2.0

This toolbox was conceived to foster musculoskeletal simulations for a large amount of data: it has the goal to decrease both processing and analysis time, and to move a step forward towards a standardization of the methodological approach, with the potential to encourage research collaborations and boost translation of applications into clinical practice. BOPS, indeed, allows to: (i) batch process a whole dataset using a single pipeline; (ii) track changes in processing parameters and compare results obtained with different setup combinations; and, (iii) organize results in structured folders to facilitate the retrieve and comparisons with previous or future analysis. BOPS was conceived and initially developed by the Rehabilitation Engineering Group (University of Padua, Italy) and made freely available to the research community in 2015 as a beta version (Mantoan & Reggiani 2015). The subsequent continuous process of extensive testing and refinement at the Human Movement Biomechanics Laboratory (University of Ottawa, Canada) led to the release of the current improved and stable version (2.0).

This paper describes the toolbox structure and then introduces a testing procedure with detailed results. Finally, the paper points out BOPS 2.0 key features and main advantages. Test data and results, freely available on the SimTK platform, show that BOPS is a user-friendly method that can efficaciously support musculoskeletal research teams in using OpenSim.

Methods

System overview

BOPS is designed to batch process data from musculoskeletal simulation using OpenSim. It was outlined in OpenSim 3.3 and it currently supports its most recent version (i.e.,

4.1). The tool targets a broad spectrum of researchers as it does not require any computer programming skills after the one-time setup to interface Matlab with OpenSim APIs¹⁷.

A GUI has been carefully designed to simplify the use of BOPS procedures. Six main panels provide the user with a full overview of the available functionalities and setup parameters: (i) *Settings*; (ii) *Inverse Kinematics* (IK); (iii) *Muscle Analysis* (MA); (iv) *Inverse Dynamic* (ID); (v) *Static Optimization* (SO); and (vi) *Joint Reaction Analysis* (JRA) (Figure 2 and Table 2). The GUI allows selecting multiple or single analyses (e.g., IK or JRA, only), and the processing of multiple or individual trials of a single subject at a time. In addition, the users can customize processing parameters (i.e., cutoff frequencies for filtering) and select the desired output results (i.e., the procedures to run and the plotting setups, such as variables of interest and axes labels) directly from the GUI.

Data management

The toolbox integrates perfectly with OpenSim API and promotes the organization of data in a pre-defined folder structure (Mantoan et al. 2015) (Figure 3–left). This design choice has several advantages: i) it provides a guideline to researchers in the setup of a new dataset; ii) it makes it easier to collaborate and share data among research groups; iii) it offers a method to keep a detailed track of the previously performed analyses. Indeed, in each trial folder, the setup files listing the parameter choices are included for future reference (System configuration section).

Input data loading

The input data are the same required by OpenSim, i.e., markers trajectories and force plate data (.trc and .mot file formats, respectively). MOtoNMS tool (Mantoan et al. 2015),

¹⁷

Scripting with MATLAB:
<http://simtk-confluence.stanford.edu:8080/display/OpenSim/Scripting+with+Matlab>

freely available, allows the transformation of data collected in a motion analysis laboratory into the inputs required by both BOPS and OpenSim, and automatically organizes them in the recommended folder structure (Figure 3-left). However, the input data can also be extracted directly using the proprietary motion capture system software and manually stored as required.

Figure 2. BOPS GUI. Layout of the user-friendly graphical Matlab interface available in BOPS. It allows to select which OpenSim procedures to perform sequentially, among *Inverse Kinematics*, *Inverse Dynamic*, *Muscle Analysis*, *Static Optimization*, and *Joint Reaction Analysis*, and to input both processing parameters and corresponding setup files. Please, refer to Table 2 for a detailed description of the GUI functionalities.

Table 2. Summary of BOPS GUI functionalities. Detailed description of the GUI's main panels and functionalities available through the GUI graphical elements. The Matlab function styles are specified, as well as options available for the processing and any additional related information.

Main panel		Function Style	Operation	Additional information
Settings	1	text	Folder where input data are located	Input data are .trc and .mot files
	2	pushbutton	Select the input data and OpenSim model folders	-
	3	text	OpenSim model's name	-
	4	listbox	Show all trials in the input data folder and allow to select those to process	Input data must be organized as shown in Figure 3
Inverse Kinematic (IK)	5	radiobutton	Select IK procedure	-
	6	popupmenu	Show IK templates available and select the one to use	Recommended: locate IK templates in <i>Templates/IK procedure</i>
	7	checkbox	Load an IK template located in a folder different from that recommended	-
	8	radiobutton	Plot IK results	<i>Custom option:</i> allow to select which results to plot <i>Plot all results option:</i> plot all IK angles based on IK template
	9	popupmenu	Select the x-axis variable for all plots	-
Muscle Analysis (MA)	10	radiobutton	Select MA procedure	-
	11	radiobutton	Plot MA results	-
Inverse Dynamic (ID)	12	radiobutton	Select ID procedure	-
	13	popupmenu	Show ID templates available and select the one to use	Recommended: locate ID templates in <i>Templates/ID procedure</i>
	14	checkbox	Load an ID template located in a folder different from that recommended	-
	15	edit	Customize the low-pass cut frequency	<i>Custom option:</i> allow to select which results to plot <i>Plot all results option:</i> plot all moments based on ID template
	16	radiobutton	Plot ID results	-
Static Optimization (SO)	17	radiobutton	Select SO procedure	-
	18	popupmenu	Show SO templates and select the one to use	Recommended: locate SO templates in <i>Templates/StaticOptimization</i>
	19	checkbox	Load a SO template located in a folder different from that recommended	-
	20	popupmenu	Select actuators	-
	21	checkbox	Load an Actuators file located in a folder different from that recommended	Recommended: locate Actuators files in <i>Templates/StaticOptimization</i>
	22	edit	Customize the low-pass cut frequency to SO analysis	<i>Custom option:</i> allow to select which results to plot <i>Plot all results option:</i> plot all output based on SO template Options: Force and Activation
	23	radiobutton	Plot SO results	Recommended: locate SO plot templates in <i>Templates/PlotStorage</i>
	24	popupmenu	Select the results to plot	-
	25	popupmenu	Load a SO plot template	-
	26	checkbox	Load a SO plot setup located in a folder different from that recommended	-
Joint Reaction Analysis (JRA)	27	radiobutton	Select JRA procedure	-
	28	popupmenu	Show JRA templates available and select the one to use	Recommended: locate JRA templates in <i>Templates/JRA</i>
	29	checkbox	Load a JRA template located in a folder different from that recommended	-
	30	edit	Customize the low-pass cut frequency to JRA analysis	<i>Custom option:</i> allow to select which results to plot <i>Plot all results option:</i> plot all output based on JRA template
	31	radiobutton	Plot JRA results	-

Figure 3. BOPS folders organization. The structure on the left presents the data organization suggested by the authors. All input data must be grouped in a *dynamicElaborations* folder, and subdivided in trials' folders that include both *.mot* and *.trc* files. This file organization can be easily obtained using MOtoNMS software. The OpenSim model file (*.osim*) can be automatically loaded on the BOPS GUI if located in a dedicated folder as recommended. The output data folders include processing results, figures, logs, and setup files and are

automatically created mirroring the structure of the *dynamicElaborations* folder. On the right, BOPS main folder organization is presented. The *.src* folder has the main source code while the *Templates* folder stores the setup files necessary to run the scripts.

Output data generation and storage structure

The output data consist of the results from OpenSim procedures, i.e., joint angles (IK), joint moments (ID), muscle length and moment arms (MA), muscle activations and forces (SO), and joint reaction forces (JRA). All outputs are automatically generated and stored mirroring the input folder structure (*dynamicElaboration*, *Figure 3-left*), relieving the users from having to organize the storing of many output files manually and resulting in a consistent data organization. BOPS also gives the user a high level of output customization, allowing to export additional file formats such as *.sto* file (OpenSim storage), setups, logs and text formats files (Table 2), as well as to generate output figures based on the selected variables of interest (Figure 2).

System configuration

BOPS has been designed to be a flexible and highly configurable toolbox that can fulfill the needs of different research projects without the necessity of accessing and changing the underlying source code. The user can indeed select directly from the intuitive and user-friendly interface: (i) the OpenSim procedures to run, (ii) the trials to process, (iii) the OpenSim model, scaled on the subject, to use on the simulations, (iv) the cutoff frequency for the filters, (v) the residual actuators, and (vi) the output variables to plot. When additional information is required for the processing, pre-defined setup files can be used to input: markers' names and weights, external loads (force plates), muscles and moment arms of interest, static optimization conditions and muscle reserve actuators loads, and joints of interest for contact force

processing. The file format used for the setup files is the *.xml* (eXtensible Markup Language) as it is extremely suitable for encoding parameter information that is both human and machine-readable. BOPS already provides several template files in dedicated subfolders (one per each supported OpenSim procedure) distributed with BOPS 2.0 source code (Figure 3–right). Researchers that need to define new setups based on their laboratory configuration and analysis of interest can easily create new setup files customizing the templates already available. Similar to the processing parameters that can be specified through the GUI, also the setup files are automatically loaded and can be selected on a list box menu in BOPS 2.0. However, if the user prefers to store the setup files in a different location, a popup menu will allow the selection of the custom files.

Data Processing

BOPS has already been used to process different datasets for multiple applications (Ceseracciu et al. 2015; Pizzolato et al. 2015; Catelli, Ng, et al. 2019; Catelli et al. 2020). However, to illustrate its potentialities, we analyzed data from a healthy participant (35.0 years, 78.2 kg, and 176 cm) performing five gait trials. A 10-camera capture system (200 Hz, MX-13 Vicon, Oxford, UK) recorded 3-dimensional kinematics of 45 full-body marker trajectories (Mantovani & Lamontagne 2017) and two force plates (1000 Hz, Bertec Corporation, Columbus, USA) recorded the ground reaction forces. The motion capture data were filtered and exported into *.mot* and *.trc* files (Nexus 2.8, Vicon, Oxford, UK). The simulation was performed using a generic musculoskeletal model (Catelli, Wesseling, et al. 2019) and included computation of joint angles (IK), joint moments (ID), and muscle activation (SO) during the stance phase. The JRA was also performed using resultant forces and moments acting on each articulating joint, the muscle forces, and the internal and external load applied to the model.

Two skilled users of OpenSim were asked to perform the analysis on the same data, first using the OpenSim 3.3 GUI directly, and then batch processing the data using BOPS 2.0.

Processing times were recorded starting when the software tools (OpenSim and BOPS, respectively) were opened and ending with the storing of the final results.

Results

The time required to process five gait trials using BOPS 2.0 was approximately 65% less time than when processed in OpenSim GUI (435 ± 4 s using BOPS, and 1205 ± 27 s using OpenSim GUI). Figure 4 shows the results obtained through the OpenSim GUI and exploiting BOPS 2.0. The test data used for this experiment are available at the SimTK platform (*TestData-v2.0*), together with BOPS 2.0 source code, to allow reproducibility of results and any additional testing.

Figure 4. Testing Procedure Results. Examples of inverse kinematic, inverse dynamic, static optimization and joint reaction results from BOPS (red) and OpenSim (black) during the stance phase of a gait trial. The results obtained with the two approaches fully overlap.

Discussion

The purpose of this paper was to present BOPS, a Matlab toolbox to batch process the most common OpenSim procedures. Although OpenSim is an extremely efficient tool for musculoskeletal modeling and simulation investigations, it requires user expertise to execute repetitive steps that sometimes can be boring, laborious, or high-intensive work for the user. To the authors' knowledge, software tools to speed-up and automatize this process have not been proposed yet; nevertheless, their availability may result crucial when the datasets to analyze are composed of multiple trials, different tasks, and multiple subjects, as it is usually the case. BOPS addresses the need and aims to significantly reduce data processing time and facilitate the analysis in OpenSim, thus reducing the risk of missteps, and fostering to process larger datasets in musculoskeletal modeling and simulation. Mirroring the OpenSim approach and similarly to MOtoNMS, BOPS was conceived to operate on data coming from a single subject at a time. It is provided with the plotting functionality for visual inspection of outcomes at all the intermediate stages (i.e., for all selected OpenSim procedures), combining the intent to speed up a process that would have to be done manually from OpenSim GUI, with the need to always keep a critical view on the results to assure the quality of simulations.

BOPS 2.0 is provided with a GUI that allows users to: i) load and select the trials to process and the pre-defined setup files, ii) decide the sequence of OpenSim procedures to run, iii) input cutoff frequencies for the filters of each OpenSim procedure, and iv) select variables to be plotted, directly from the intuitive interface. This makes batch data processing more straightforward and faster than processing the trials individually through the OpenSim GUI, as also confirmed by the time required to elaborate the Test Data (approx. 65% shorter). It can reasonably be expected that the time gain would be much more appreciable as the dataset dimension and the number of trials increase.

The customizable selection of all parameters of interest on the single GUI layout offered by BOPS 2.0 has the potential not only to speed up the configuration process but also to avoid incorrect selection or inaccurate manual inputs on the numerous panels when executing multiple times the same analyses using the OpenSim GUI.

Since BOPS relies on OpenSim APIs, fully matching results were expected among the two different approaches, which were indeed obtained (Figure 4), confirming that BOPS can be considered a reliable tool.

Moreover, it is always possible to load the setup or output files generated with BOPS into OpenSim, for results reproduction or visualization. BOPS, indeed, stores output files and figures, and processing logs and .xml setups files into output directories automatically generated and organized with a uniquely defined structure. This standard approach to dataset organization facilitates data exchange among research groups, which can customize their tools and applications relying on a common data structure.

Results show that BOPS 2.0 succeeds in processing musculoskeletal data consistently and repeatably based on the user-defined system configuration. The Test Data used in this study are included as a supplementary file for results replicability, while additional test data are available on the SimTK project platform (*TestData-v0.9*) to allow the testing of the tool on different datasets. A tutorial video is also available in the BOPS project on the SimTK platform to illustrate how to use the toolbox and to provide an overview of its potentialities.

BOPS is released under Apache License 2.0 to encourage its use and it is freely available on the SimTK platform (<https://simtk.org/projects/bops>). The latest updates can be found on the GitHub repository (<https://github.com/RehabEngGroup/BOPS>), where the OpenSim research community is strongly encouraged to collaborate on its improvement. Authors' plans for future developments include the addition of other OpenSim procedures (i.e., Computed Muscle Control) and the improvement of real-time feedback during the processing,

logging files, data visualization and output plotting features. This could allow to evaluate the implementation of an additional layer to process multiple subjects as a major update in the future, thus further speeding up the processing of large datasets while limiting the risk of compromising accuracy and reliability of results.

BOPS has been developed in Matlab for its widespread use in the biomechanics field (Table 1). Other freely available development platforms, such as Python or GNU Octave were considered five years ago, during the initial stages of this project, but discarded as potentially less comfortable for the users working on this field. Indeed, when directly compared with Python (Python Software Foundation), the PubMed database show a higher incidence of “biomechanics toolbox” and “human movement toolbox” wording associated with Matlab (13 and 41, respectively) instead of Python (two and seven, respectively). However, Python scripting and extensions of the OpenSim API is available since OpenSim 3.2 and other biomechanics software are being provided with a Python package and/or binding (e.g., BTK, since 2014). Therefore, for its increasing use also in this field, for the higher computing performance, and to encourage adoption also from users who could prefer a free option, a Python BOPS version will be implemented in the near future.

The authors expect that BOPS will be useful to the research community by significantly reducing the data processing time, facilitating OpenSim analysis, and boosting research group collaborations. The availability of such an easy-to-use and highly configurable toolbox will potentially facilitate researchers to use musculoskeletal modeling and simulations in research, training, and clinical applications.

Funding

This study was partially financed by the Coordenação de Aperfeiçoamento de Pessoal de Nível Superior – Brasil (CAPES) [Finance Code 001]; Canadian Institutes of Health Research (CIHR) [PJT-162333].

Availability and Requirements

Project name: BOPS

Project link: SimTK (<https://simtk.org/projects/bops>)

GitHub (<https://github.com/RehabEngGroup/BOPS>)

Programming language: Matlab

License: Batch OpenSim Processing Scripts (BOPS) Licensed under the Apache License, Version 2.0 (the "License")

Any restrictions to use by non-academics: None

Software requirements: OpenSim 3.3 and 4.x.

Supplementary files

Test Data v2.0 ([Link](#))¹⁸

Tutorial video exemplifying how to use BOPS ([Link](#))¹⁹.

Disclosure statement The authors do not have any conflicts to declare that could inappropriately influence this article.

¹⁸ Em caso de problemas em acessar o arquivo, segure a tecla “ctrl” e clique no link abaixo:

Link Test Data v2.0:

https://simtk.org/account/login.php?triggered=1&return_to=%2Ffrs%2Fdownload_confirm.php%2Ffile%2F6215%2FTestData-2.0.zip%3Fgroup_id%3D1038

¹⁹ Link Tutorial Vídeo: https://simtk.org/docman/?group_id=1038

Se o problema persistir, por favor envie um e-mail para: bruno.bedo@usp.br ou paulosantiag@usp.br

References

- Barre A, Armand S. Biomechanical toolkit: Open-source framework to visualize and process biomechanical data. *Comput Methods Prog Biomed*. 2014; 114(1):80–7.
- Bedo BLS, Catelli DS, Lamontagne M, Santiago PRP. 2020. A custom musculoskeletal model for estimation of medial and lateral tibiofemoral contact forces during tasks with high knee and hip flexions. *Comput Methods Biomech Biomed Engin*. 23(10):658–663.
- Catelli DS, Ng KCG, Kowalski E, Beaulé PE, Lamontagne M. 2019. Modified gait patterns due to cam FAI syndrome remain unchanged after surgery. *Gait Posture*. 72:135–141.
- Catelli DS, Ng KCG, Wesseling M, Kowalski E, Jonkers I, Beaulé PE, Lamontagne M. 2020. Hip Muscle Forces and Contact Loading During Squatting After Cam-Type FAI Surgery. *J Bone Jt Surg*. 102(Suppl 2):34–42.
- Catelli DS, Wesseling M, Jonkers I, Lamontagne M. 2019. A Musculoskeletal Model Customized for Squatting Task. *Comput Methods Biomech Biomed Engin*. 22(1):21–24.
- Ceseracciu E, Mantoan A, Bassa M, Moreno JC, Pons JL, Prieto GA, Del Ama AJ, Marquez-Sanchez E, Gil-Agudo A, Pizzolato C, et al. 2015. A flexible architecture to enhance wearable robots: Integration of EMG-informed models. In: *IEEE/RSJ Int Conf Intell Robot Syst*. Hamburg; p. 4368–74.
- Damsgaard M, Rasmussen J, Christensen ST, Surma E, de Zee M. 2006. Analysis of musculoskeletal systems in the AnyBody Modeling System. *Simul Model Pract Theory*. 14(8):1100–1111.
- Davoodi R, Brown IE, Loeb GE. 2003. Advanced modeling environment for developing and testing FES control systems. *Med Eng Phys*. 25(1):3–9.
- Delp SL, Anderson FC, Arnold AS, Loan P, Habib A, John CT, Guendelman E, Thelen DG. 2007. OpenSim: Open-Source Software to Create and Analyze Dynamic Simulations of Movement. *IEEE Trans Biomed Eng*. 54(11):1940–1950.
- James DA, Wixted A. 2011. ADAT: A Matlab toolbox for handling time series athlete performance data. *Procedia Eng*. 13:451–6.
- Lerner ZF, DeMers MS, Delp SL, Browning RC. 2015. How tibiofemoral alignment and contact locations affect predictions of medial and lateral tibiofemoral contact forces. *J Biomech*. 48(4):644–650.
- Mantoan A, Pizzolato C, Sartori M, Sawacha Z, Cobelli C, Reggiani M. 2015. MOtoNMS: A MATLAB toolbox to process motion data for neuromusculoskeletal modeling and simulation. *Source Code Biol Med*. 10(1):12.

- Mantoan A, Reggiani M. 2015. BOPS v0.9 (Version v0.9). Zenodo.
- Mantovani G, Lamontagne M. 2017. How different marker sets affect joint angles in inverse kinematics framework. *J Biomech Eng.* 139(4):044503.
- Martinez R, Michaud B, Begon M. 2020. pyomeca: An Open-Source Framework for Biomechanical Analysis. *J Open Source Softw.* 5(53):2431.
- Pizzolato C, Lloyd DG, Sartori M, Ceseracciu E, Besier TF, Fregly BJ, Reggiani M. 2015. CEINMS: A toolbox to investigate the influence of different neural control solutions on the prediction of muscle excitation and joint moments during dynamic motor tasks. *J Biomech.* 48(14):3929–36.
- Rajagopal A, Dembia CL, DeMers MS, Delp DD, Hicks JL, Delp SL. 2016. Full body musculoskeletal model for muscle-driven simulation of human gait. *IEEE Trans Biomed Eng.* 63(10):2068–79.
- Seth A, Hicks JL, Uchida TK, Habib A, Dembia CL, Dunne JJ, Ong CF, DeMers MS, Rajagopal A, Millard M, et al. 2018. OpenSim: Simulating musculoskeletal dynamics and neuromuscular control to study human and animal movement. *PLOS Comput Biol.* 14(7):e1006223.
- Zhang J, Sorby H, Clement J, Thomas CDL, Hunter P, Nielsen P, Lloyd D, Taylor M, Besier T. 2014. The MAP client: User-friendly musculoskeletal modelling workflows. In: Bello F, Cotin S, editors. *Biomed Simul.* Vol. 8789. Strasbourg, France: Springer; p. 182–192.

Table A1. Main functionalities of Open Source toolboxes based on OpenSim for musculoskeletal modeling and simulations

Musculoskeletal Modeling & Simulation Toolboxes based on OpenSim	Programming Language	Main Functions
BOPS 2.0	Matlab	<ul style="list-style-type: none"> • GUI • Data Filtering • Customize the low-pass cut frequency • OpenSim Inverse Kinematics (IK) • OpenSim Inverse Dynamics (ID) • OpenSim Muscle Analysis (MA) • OpenSim Static Optimization (SO) • OpenSim Joint Reaction Analysis (JRA) • Load Templates for IK, ID, SO, JRA • Load Actuators • Store setup files • Results Visualization • Plot Customization • Standardized dataset organization • Open-source
Mansouri & Reinbolt (2012)	Matlab/Simulink	<ul style="list-style-type: none"> • Forward Dynamics simulation of OpenSim models • Design and application of open-loop and closed-loop control systems • Real-time changes to input controls (feedback control) • Private access

CAPÍTULO 6

ESTUDO 5

Effect of the fatigue in knee biomechanics during the sidestep cutting maneuver: a modeling approach.

Bruno Luiz Souza Bedo; Danilo S. Catelli; Mario Lamontagne; Renato Moraes; Paulo Roberto Pereira Santiago.

Effect of the fatigue in knee biomechanics during the sidestep cutting maneuver: a modeling approach.

Bruno Luiz Souza Bedo^{ABC}; Danilo Santos Catelli^B; Mario Lamontagne^B; Renato Moraes^{AC}; Paulo Roberto Pereira Santiago^{AC}

^ARibeirão Preto Medical School, University of São Paulo, Ribeirão Preto, Brazil.

^BFaculty of Health Sciences, University of Ottawa, Ottawa, Canada.

^CSchool of Physical Education and Sports of Ribeirão Preto, University of São Paulo, Ribeirão Preto, Brazil.

Observação:

O presente estudo encontra-se em processo de revisão.

Abstract

Knee injuries are frequently observed during non-contact situations and fatigue has been suggested to be related to the decrease in joint stability and compromises the stability of the ligaments; however, there is no consensus that this argument is fully correct. Hence, it becomes an important methodological decision to investigate the effect of fatigue on musculoskeletal dynamics during injury risk maneuvers. Although musculoskeletal models provide a non-invasive approach to analyze movement, many studies have examined the tibiofemoral contact forces considered calculating the tibiofemoral load at a single point in the joint, so the effect of the fatigue in the lateral and medial compartments during sidestep cutting still lacking. Therefore, the present proposal aims to investigate the effect of a specific fatigue protocol for handball athletes on knee biomechanics during sidestep cutting maneuver using a modeling approach. The modeling approach allowed to estimate of tibiofemoral contact forces in the medial and lateral components, as well the relative contributions of individual lower-limb muscles during the sidestep cutting. Twenty female handball athletes performed three trials of sidestep cutting maneuver (SCM) before (baseline) and after the fatigue protocol. Eight retroreflective-cameras (Vicon, UK) and one force plate (Bertec, US) tracked the marker trajectories and measured the ground reaction forces, respectively. The data were modeled using the OpenSim where a subject-specific model will be used to calculate joint angles, moments, muscles, and tibiofemoral contact forces. The variables were time-normalized, and their waveforms were statistically compared using the statistical parametric mapping (SPM), with a significance level of $p < 0.05$. The main results pointed to a decrease in knee flexion in the fatigued condition during the first instants of the stance phase. Moreover, post-fatigue protocol analysis also demonstrated significantly reduced vasti muscle force production. Similarly, the SPM analysis showed a decrease in the tibiofemoral contact forces in the vertical and anterior. Lastly, fatigue demonstrated significantly reduced both medial and lateral force components. The results indicate that forces applied to the tibiofemoral joint are reduced when fatigued compared to the baseline values. The results reinforce that there is no consistent evidence that fatigue increases the risk of knee injuries.

1. Introduction

Injuries in the knee joint, especially noncontact anterior cruciate ligament (ACL) ruptures, are frequently observed during single- and double-leg landings, change of direction and pivot movements (Andrews, McLeod, Ward, & Howard, 1977; Bencke et al., 2012; Boden et al., 2000; Hewett et al., 2001; Koga et al., 2010; B. Yu & Garrett, 2007). In handball, as in other team sports, such as soccer and basketball, ACL injuries are common, especially regarding young female athletes (Bedo et al., 2019; Bencke et al., 2012; Langevoort et al., 2007; Myklebust, Maehlum, Holm, & Bahr, 1998; Olsen et al., 2004, 2006; Piry et al., 2011). The ruptures usually occur during landing maneuvers or sidestep cutting with a slight knee flexion associated with knee valgus and internal rotation (Alentorn-Geli et al., 2009; Hewett et al., 2009). Due to a large number of athletic injuries observed during sidestep cutting maneuvers, different groups have investigated lower limb joint biomechanics of such task in a laboratory environment (Bedo, Cesar, et al., 2021; Guilherme M Cesar et al., 2011; Dempsey, Lloyd, Elliott, Steele, & Munro, 2009; Kristianslund et al., 2014; M. J. C. Lee, Lloyd, Lay, Bourke, & Alderson, 2016; McLean et al., 2005; Mok, Bahr, & Krosshaug, 2018; Monfort et al., 2019; Sigward et al., 2015), as well as the intrinsic risk factors for injuries (Posthumus, Collins, September, & Schwellnus, 2011).

The impact of fatigue on knee biomechanics has been frequently discussed in the literature. Studies have related muscle fatigue to an increased risk of ACL injury (Borotikar et al., 2008; Liederbach et al., 2014; McLean et al., 2007; Xia et al., 2017); however, there is no consensus that this argument is fully correct (Bourne et al., 2019). In theory, when fatigued, the muscles absorb less energy than when non-fatigued (Mair et al., 1996) and as a consequence, deleterious alterations in the lower limb joint biomechanics can increase the risk of injury. The researchers based these statements on the theory that fatigue results in a proprioceptive deficit and delayed muscle response (Hiemstra et al., 2001; Lattanzio et al., 1997). However, epidemiologic studies have not shown a consistent connection between injury

and fatigue, as a result of playing time (Waldén et al., 2011). Although some studies defend fatigue increases the risk of injuries, a recent review pointed that fatigue has no significant influence on most of the knee kinematic and kinetic variables (Benjaminse et al., 2019). Further, they highlighted that few are the studies that quantified fatigue as a risk factor and that there is no evidence of fatigue protocols that significantly impacted known variables to increase the risk of injury. Moreover, to the best of our knowledge, the majority of studies that investigated the effect of fatigue in lower limb biomechanics used usual biomechanical approaches, such as kinematics and kinetics. Therefore, it becomes an important methodological decision to investigate the effect of the fatigue phenomenon and their interactions of neural control with musculoskeletal dynamics during injury risk maneuvers.

Methods that measure knee contact forces *in vivo* are extremely invasive and expensive (Bergmann, Graichen, & Rohlmann, 1993; Fregly et al., 2012); hence, musculoskeletal models provide a non-invasive approach to analyze movement. For instance, the OpenSimTM (Stanford University, USA), an open-source software with a huge community of researchers, enables computational modeling of musculoskeletal anatomy and dynamic functions (Delp et al., 2007). As an example of the possibility of analysis, OpenSim enables the calculation of joint contact forces on different joints and different tasks (Catelli, Wesseling, et al., 2019a; Lai et al., 2017). Therefore, the implementation of computation modeling becomes a powerful tool to understand the effect of fatigue on the knee joint.

The tibiofemoral joint endures significant mechanical loads during different tasks and the asymmetry on the medial and lateral alignment impacts the load redistribution on the condyles, increasing the risks of osteoarthritis (Jackson et al., 2004). Moreover, the compressive forces are considered to be important for optimizing the joint stability during sidestepping (Killen et al., 2018). However, many studies have examined the tibiofemoral contact forces considered calculating the tibiofemoral load at a single point in the joint. Up to

now, only two studies (Killen et al., 2018; Saxby et al., 2016) have investigated the tibiofemoral contact forces in the medial and lateral components during the sidestep cutting maneuver. The effect of the fatigue in the lateral and medial compartments during sidestep cutting still lacking.

Therefore, the purpose of this study was to evaluate the effect of a handball-specific fatigue protocol in the estimation of tibiofemoral contact forces in the medial and lateral components, as well the relative contributions of individual lower-limb muscles during the sidestep cutting maneuver in female handball athletes. We hypothesized that the lower limb muscle and tibiofemoral contact forces will reduce when the fatigue state is reached.

2. Objective

2.1 Participants

Twenty collegiate female handball athletes (1.76 ± 0.07 m; 63.5 ± 9.1 kg; 21.9 ± 3.4 years) volunteered for this study. All participants had no history of knee injuries, as well as lower limb injuries in the past six months. They also had long experience with handball practice (7.2 ± 3.2 years of experience playing handball) and during data collection, all athletes were at the beginning of the competitive season (240 ± 180.4 minutes of practice/week). Goalkeepers were excluded from the sample because the sidestep cutting maneuver is not a task that they usually perform. All participants were right-handed, and the lower limb dominance was determined by asking the leg used to take off in a handball jump shot (Zech et al., 2012). All procedures were approved by the local Research Ethics Committee and all participants provided written consent.

2.2 Methods

2.2.1 Fatigue protocol

All procedures were performed at the biomechanics laboratory, including the fatigue protocol, which was composed of tasks performed during a handball practice. These tasks, such as lateral and back shifting, sidestep cutting, jumping and landings, among others, were

performed in a circuit format (Bedo, Pereira, et al. (2020)), where they should perform as fast as possible, and the protocol was divided into rounds, with a gradual increase of laps (i.e., round I [1 lap]; round II [2 laps]; round III [3 laps]; and so on).

To quantify the effect of the intensity of the protocol, the effort time, heart rate (HR, Polar V800™) and rate of perceived exertion (RPE using the Foster scale [0–10](Foster et al., 2001)) were assessed at the end of each lap. Also, once the fatigue is characterized by the inability to produce force, at the end of each round, the maximum propulsion force was quantified by a countermovement jump performed on a force plate. The participants were instructed to perform the countermovement jump as high as possible without flexing their knees or hips during the flight phase. The fatigued condition was characterized by the participant's inability to continue performing the protocol.

2.2.2 Motion Analysis

To quantify the impact of the fatigue on the lower limb biomechanics, the participants performed six valid trials of sidestep cutting maneuvers (SCM) with the non-dominant leg: three before and three immediately after the fatigue protocol. The lower limb dominance was determined by asking the athletes about the leg they used to take off in the handball shoot (Zech et al., 2012), and as all participants were right-leg dominant, they all performed with the left leg. To perform the SCM, the athletes should sprint for four meters and perform a 45° cutting maneuver from the original movement. A color tape was placed on the floor to indicate the direction of cutting movement, and a valid trial occurred when the participant remained over the tape and all athletes wore their handball shoes during all data collection. Regarding the tasks after exhaustion, the tasks were performed during the first five minutes after the fatigue detection, while the fatigue effects were still present (Bedo, Pereira, et al., 2020).

The kinematics were collected during the sidestep cutting maneuvers by a motion capture system with 8 infrared cameras (250 Hz, MX-T40S; Vicon, Oxford, United Kingdom),

that tracked passive retro-reflective markers (10 mm diameter) according to the University of Ottawa Motion Analysis Model (UOMAM) marker set (Mantovani & Lamontagne, 2017a). In addition, a first force plate (40 × 60 cm, Bertec, Columbus, USA, sampled at 2000 Hz) quantified the force and moment data. The marker trajectories were labeled and all data were filtered (zero-lag, fourth-order Butterworth filter, cut-off frequency = 6 Hz) and exported into .mot and .trc files in Nexus 2.8 (Vicon, Oxford, UK).

A second force plate was used to calculate the perceptual change in maximum propulsion force production (F_{MAX}) during the countermovement jump performed before and immediately after the fatigue protocol. The data was normalized by the participants' total body weight and the F_{MAX} was determined as the first peak of the ground reaction force. A Matlab® script (Mathworks Inc. Natick, MA, USA) collected and processed the force data, with a 10-Hz low-pass fourth-order Butterworth filter.

2.4 Musculoskeletal modeling

The generic musculoskeletal model used was composed of 80 lower limb Hill-type muscle-tendon units, 17 torque actuators driving the upper body and 37 degrees of freedom. In addition, the model also included medial and lateral knee compartment geometries, allowing the quantification of the vertical tibiofemoral contact force (Bedo, Catelli, et al., 2020). The modeling analysis were all processed in OpenSim™ 3.3 (Stanford University, Stanford, USA) (Delp et al., 2007) while using the BOPS toolbox for data batch processing (Inverse Kinematics, Inverse Dynamics, Muscle Analysis, Static Optimization, and Joint Reaction Analysis) in Matlab (Bedo, Mantoan, et al., 2021). Regarding specifically the joint reaction analysis on the tibiofemoral joint, the musculoskeletal model allowed quantifying the global contact force in the anteroposterior, mediolateral and vertical direction. In addition, the model calculated the vertical force of the medial and lateral condyles, allowing a better comprehension of the effect of the fatigue on the knee condyles.

2.4 Statistical analysis

The values of F_{MAX} , heart rate and rate of perceived exertion before and after the fatigue protocol were compared using paired samples t-tests using the SPSS Statistics v.23 (IBM Corporation, Armonk, USA). Furthermore, to quantify the impact of the fatigue on biomechanics variables, the results analyzed during the stance phase (between foot-strike and foot-off) of the SCM were time-normalized (0-100%). Hence, the data were compared by paired analysis using statistical parametric mapping (SPM) (Pataky, 2010) with the significance level set at $p < 0.05$.

3. Results

3.1 Fatigue parameters

The athletes reduced their F_{MAX} by approximately ~10% in the fatigue situation when comparing to baseline. In contrast, the heart rate, effort time and rate of perceived exertion increased significantly after the fatigue protocol (Table 1).

Table 1. Mean and standard deviation of the fatigue parameters.

	Baseline	Fatigue
Heart rate (bpm) *	82.45 ± 8.14	194.35 ± 7.38
F_{MAX} (BW)*	2.51 ± 0.51	2.24 ± 0.30
Rate of Perceived Exertion *	0 ± 0	10 ± 0
Effort time (s)*	42.12 ± 4.95 (first lap)	58.85 ± 9.39

* $p < 0.0001$

3.2 Joint kinematics and kinetics

SPM analysis showed a decrease in knee flexion in the fatigued condition during the first instants of the stance-phase (0-20% of the cycle; $p = .04$; $t^* = 2.6$; Figure 1). There was no difference in the other lower limb joints kinematics. No differences were found in the joint moments.

Figure 1. Knee flexion angle during sidestep cutting maneuver. The black and red lines represent the average angles for the baseline and fatigue conditions, respectively. The corresponding colors of the shaded areas represent the standard deviation of each condition. The black bars with red striped denote the significant differences areas indicated by the SPM analysis.

3.3 Muscle and tibiofemoral contact forces

Post-fatigue protocol analysis significantly reduced vastus intermedius (3-29% of the cycle; $p < .001$; $t^* = 3.2$) and vastus medialis (4-22% of the cycle; $p < .001$; $t^* = 3.2$) immediately after ground contact. During the propulsion phase ($> 50\%$ of the cycle), the vastus intermedius (60-96% of the cycle; $p < .0001$; $t^* = 3.2$), vastus lateralis (88-93% of the cycle; $p = 0.26$; $t^* = 3.1$) and vastus medialis (69-95% of the cycle; $p = 0.02$; $t^* = 3.2$) had their forces significantly reduced when fatigued (Figure 2).

There were no differences between baseline and fatigue values in the mediolateral directions of the tibiofemoral contact forces. Conversely, the SPM analysis showed a decrease in the tibiofemoral contact forces in the vertical (15-28% and 57-64% of the cycle; $p = 0.0009$ and 0.01 , respectively; $t^* = 3.2$) and anterior (15-43% of the cycle; $p < 0.0001$; $t^* = 3.1$; Figure 3).

Lastly, regarding the medial and lateral condyles, the fatigue demonstrated significantly reduced both medial (15-43% of the cycle; $p = 0.01$) and lateral (53-65% of the cycle; $p = 0.001$) force components (Figure 4).

Figure 2. Vastus muscles forces during sidestep cutting maneuver task. The red and black lines represent the average angles for the fatigue and baseline conditions, respectively. The black and red lines represent the average muscle forces for the baseline and fatigue conditions, respectively. The corresponding colors of the shaded areas represent the standard deviation of each condition. The black bars with red striped denote the significant differences areas indicated by the SPM analysis.

Figure 3. Tibiofemoral contact forces on anterior-posterior (A), superior-inferior (B) and medial-lateral (C) directions during sidestep cutting maneuver task. The red and black lines represent the average of tibiofemoral forces for the fatigue and baseline conditions, respectively. The corresponding colors of the shaded areas represent the standard deviation of each condition. The black bars with red striped denote the significant differences areas indicated by the SPM analysis.

Figure 4. Tibiofemoral contact forces during the sidestep cutting maneuver. The graphs on the left column represent the medial (A) and lateral (B) components' reaction forces during the task. The red and black lines represent the average of tibiofemoral forces for the fatigue and baseline conditions, respectively. Lastly, the graphs on the right column show the body weight (BW) normalized tibiofemoral contact forces resultant during the task, reported with the compass graph on lateral (C) and posterior view (D). The red and black lines show the magnitude (radar graph) and direction to the tibial plateau obtained from the musculoskeletal model.

5. Discussion

The purpose of this study was to investigate the effect the fatigue on knee biomechanics, using a musculoskeletal approach, during the sidestep cutting maneuver task handball athletes. Besides the usual biomechanical variables, such as joint angles and moments, the musculoskeletal model allowed estimation of muscle forces and tibiofemoral contact forces in

the medial and lateral condyles. The main results demonstrated that the fatigue protocol impacted knee load. More specifically, we observed a decrease in the knee joint flexion after the fatigue protocol. The fatigued state significantly reduced the vasti muscles forces and the tibiofemoral contact forces in the vertical and anterior directions, as well the forces on the medial and lateral components. These results provide insights into how fatigue impacts knee joint loading.

Fatigue is characterized as an exercise-induced reduction in the ability to produce force; hence, the decrease in the F_{MAX} and muscle forces, indicates that the specific fatigue protocol was effective to induce fatigue among the handball athletes. This protocol has already shown efficiency to impact postural control (Bedo, Pereira, et al., 2020), and the current findings support the statement that fatigue impacts lower limb biomechanics (Borotikar et al., 2008), especially when the knee presents smaller flexion angles (Chappell et al., 2005). Specifically, the athletes performed the sidestep cutting maneuver with a lower knee flexion compared to baseline values during the landing preparation (weight-acceptance period, defined as first instants after the initial ground contact). The weight-acceptance period can be determinant for an injury (Koga et al., 2010) when performing the task with lower knee flexion (Koga et al., 2010; Laughlin et al., 2011; Makaruk et al., 2014; Wang, 2011). The decrease in the knee flexion impacts the patellar tendon-tibial shaft and the ACL elevation angle (angle of hamstrings' insertion) and may increase the knee load (Bing Yu, Lin, & Garrett, 2006). Lower flexion angles limit the hamstring's potential to counteract the anterior tibial strain when landing during the sidestep cutting maneuver (B. Yu & Garrett, 2007). However, although the sagittal plane mechanism has been discussed for non-contact knee injuries (Chappell, Yu, Kirkendall, & Garrett, 2002; Griffin et al., 2000), the sagittal plane forces applied to the knee joint during sidestep cutting maneuver is not the main factor for a knee injury (McLean, Huang, Su, & van den Bogert, 2004). The movement needs to combine a lower knee flexion angle and

a large quadriceps force (Dürselen, Claes, & Kiefer, 1995). Therefore, our analysis showed that the fatigue significantly reduced the vastus intermedius and vastus medialis immediately after ground contact, which may have reduced the anterior tibial strain even with the low knee flexion. These findings may provide the first insights that fatigue is not the main factor for knee injury occurrence.

Peripheral fatigue is characterized by a failure in the contracting muscle (Fitts, 1994) and the central fatigue occurs with a considerable reduction of the neural drive to the muscles (Gandevia, 2001); therefore, neuromuscular fatigue combines central and peripheral processes (Martin et al., 2010). The protocol-induced fatigue has the potential to cause changes in different systems and may induce neuromuscular fatigue, representing the performance during sports practice, where high-intensity efforts are often required for extended periods of time and also reactive movements during interactions with opponents. But independently of the fatigue's origin, the vasti muscles had their force production reduced when fatigued, and these changes may have impacted the position matching ability (Givoni, Pham, Allen, & Proske, 2007; Hiemstra et al., 2001). Therefore, although different studies have hypothesized that fatigue reduces the muscle forces and consequently result in deleterious alterations in lower limb joint kinematics and kinetics (Benjaminse et al., 2008; Wojtys, Beaulieu, & Ashton-Miller, 2016), which may serve to increase strain on the knee joint load, the reduction on the vastus muscle forces culminates in a decrease on the tibiofemoral contact forces on the anterior-posterior direction.

The decrease in the tibiofemoral contact forces on the anteroposterior direction may result in a reduction in the shear force in the tibia and, consequently, a decrease in the knee joint ligament load. Nevertheless, the relationship between the risk of knee injuries and ground vertical forces has been previously reported (Hewett & Myer, 2011; Kristianslund & Krosshaug, 2013; Weinhandl et al., 2013; B. Yu & Garrett, 2007; Zahradnik, Uchytel, Farana,

& Jandacka, 2014). Therefore, the combination of the reduction in both anteroposterior and vertical tibiofemoral (including the medial e lateral components) contact forces reinforces the statement that fatigue does not cause knee injuries (Bourne et al., 2019). In addition, although no differences were found in the tibiofemoral contact force in the mediolateral direction, we think it is important to point out that in both conditions (baseline and fatigue) the force direction was predominantly medial. This finding strengthens that during the sidestep cutting maneuver, the knee joint load suffers an increase of the load in the soft tissues, and a possible increase in mediolateral load (frontal plane) has been previously discussed as one of the main factors to the ACL injury (G. M. Cesar et al., 2011; Dempsey et al., 2009; Herrington, 2011; Kristianslund, Faul, Bahr, Myklebust, & Krosshaug, 2012; Ludwig, Simon, Piret, & Becker, 2017; McLean et al., 2005; McLean, Huang, et al., 2004; Monfort et al., 2019; Palmieri-Smith, McLean, Ashton-Miller, & Wojtys, 2009; Quatman & Hewett, 2009; B. Yu & Garrett, 2007; Zahradnik et al., 2014).

The muscle forces are fundamental to joint stability during dynamic tasks. Regarding the sidestep cutting maneuver, during the weight-acceptance period, the agonistics muscles are responsible to increase proximal tibial anterior shear forces. On the other hand, the antagonistic response is believed to limit anterior tibial translation and, consequently, reduce the risk of injury. Therefore, the quadriceps and hamstrings co-contraction provide dynamic stability at the knee joint (David G Lloyd & Buchanan, 2001). Although the fatigue protocol has been elaborated to general fatigue, only the vasti muscles were impacted. Interestingly though, all muscles that fatigued demonstrated significantly reduced force production, and are mono-articular. Differently, no bi-articular muscles, including rectus-femoris and hamstrings, reduced their force production. There are different neuronal coding mechanisms for the bi- and mono-articular muscles (Ebenbichler et al., 1998). While the main function of mono-articular muscles is to generate force, the bi-articular muscles are responsible for controlling net torque

at the hip and knee joints and to transfer power between joints (Hautier et al., 2000; Jacobs & van Ingen Schenau, 1992). Therefore, these statements likely explain the reduction in the F_{MAX} and in the muscle forces production.

In addition, during the weight-acceptance period of the sidestep cutting maneuver, as well as during the tasks performed during the fatigue protocol (i.e. double-leg plyometric jumps and double-leg lateral jumping/landing) the participants showed a higher than 50° hip flexion. However, the rectus femoris is mainly used during 0°–45° hip flexion while the rectus abdominal activation is used for hip flexion at 45° and greater (S. Y. Lee, 2015). Also, the rectus femoris does not contribute significantly to knee extension strength since its action as a knee extensor is limited when the hip is flexed (Hamill & Knutzen, 2006). Therefore, we suggest that the rectus femoris were lower required during instances of high force production, such as the weight-acceptance period. Lastly, the difference between mono- and bi-articular muscle force can also be an effect of evidence of a principle of motor control. Motor neurons that innervate bi-articular muscles appear to receive complex mixtures of multimodal inputs. On the other hand, the motor neuron responses evoked through stimulation of hindlimb afferents demonstrated that motor neurons of mono-articular muscles show excitatory/inhibitory responses (Doorenbosch, Welter, & van Ingen Schenau, 1997; Ebenbichler et al., 1998; Jacobs & van Ingen Schenau, 1992). As an example, in the case of the rectus-femoris, a bi-articular muscle has two neurovascular branches originating from the femoral nerve, meaning that each region can be regulated individually by the central nervous system (Akima, Takahashi, Kuno, & Katsuta, 2004; Yang & Morris, 1999), while the vasti muscles have only one. Therefore, although we have not specified the type of fatigue induced by the protocol, the rectus femoris could have been less impacted by the fatigue protocol compared to the vasti muscles. Hence, future studies must investigate the different effects of

general fatigue in the different portions of the quadriceps muscle and how it affects motor control.

Some limitations affected this study. First, all the procedures, including the fatigue protocol, were performed in a laboratory environment. Aiming at an ecological approach, future studies should be conducted on the court or using real games to induce fatigue. Second, the muscle forces were calculated using a static optimization approach. Although this method has the advantage of not requiring invasive access and has been validated to various walking speeds (Modenese & Phillips, 2012), it has been showing a sensitive optimization method to calculated muscle forces and joint contact forces (Wesseling et al., 2015). However, future studies must include electromyographic data as an input variable to estimated muscle and tibiofemoral contact forces (Gerus et al., 2013; D. G. Lloyd & Besier, 2003; Wesseling et al., 2015). Lastly, the current neuromusculoskeletal model presents a 1 degree of freedom in the knee joint. That was chosen because the tibiofemoral contact forces are very sensitive to the alignment of the lower-limb joints (Jackson et al., 2004; Lerner et al., 2015), so a possible non-neutral alignment could result in inaccurate predictions of knee contact forces. But as we have not exposed our participants to imaging scans, we could not parameterize the participants' lower-limb alignment. Future studies should include imaging scans to standardize the lower-limb alignment as well as use models that allow quantifying the joint kinematic and kinetics in the frontal and transverse planes.

6. Conclusion

To the best of our knowledge, this is the first study that evaluates the fatigue effects during sidestep maneuver using a modeling approach. The fatigue impacted the knee biomechanics, reducing vasti muscle force production, as well the tibiofemoral contact force in the anteroposterior and vertical directions and over the lateral and medial condyles. These findings indicate that forces applied to the tibiofemoral joint are reduced when fatigued. Still,

there is no consistent evidence that fatigue increases the risk of knee injuries. Future studies should be performed for a better understanding of the fatigue phenomenon on lower limb biomechanics.

7. Acknowledgments

The authors thank the technical staff and athletes for participating and contributing to the current study.

8. Funding

This study was financed in part by the Coordenação de Aperfeiçoamento de Pessoal de Nível Superior – Brasil (CAPES) – Finance Code 001; by the São Paulo Research Foundation – FAPESP, Brazil [grants 2018/20362-8 and 2019/04721-0]; a financial contribution from the Human Movement Biomechanics Laboratory at the University of Ottawa – Canada; and also from the Brazilian Society of Biomechanics.

References

- Akima H, Takahashi H, Kuno SY, Katsuta S. 2004. Coactivation pattern in human quadriceps during isokinetic knee-extension by muscle functional MRI. *Eur J Appl Physiol.* 91(1):7-14.
- Alentorn-Geli E, Myer GD, Silvers HJ, Samitier G, Romero D, Lázaro-Haro C, Cugat R. 2009. Prevention of non-contact anterior cruciate ligament injuries in soccer players. Part 1: Mechanisms of injury and underlying risk factors. *Knee surgery, sports traumatology, arthroscopy.* 17(7):705-729.
- Andrews JR, McLeod WD, Ward T, Howard K. 1977. The cutting mechanism. *The American journal of sports medicine.* 5(3):111-121.
- Bedo BLS, Catelli DS, Lamontagne M, Santiago PRP. 2020. A custom musculoskeletal model for estimation of medial and lateral tibiofemoral contact forces during tasks with high knee and hip flexions. *Computer methods in biomechanics and biomedical engineering.* (in press).
- Bedo BLS, Cesar GM, Moraes R, Mariano FP, Vieira LHP, Andrade VL, Santiago PRP. 2021. Influence of Side Uncertainty on Knee Kinematics of Female Handball Athletes During Sidestep Cutting Maneuvers. *J Appl Biomech.* 1-8.
- Bedo BLS, Manechini JPV, Nunomura M, Menezes RP, Silva SRD. 2019. Injury Frequency in Handball Players: A Descriptive Study of Injury Pattern in São Paulo State Regional Teams. *Motriz: Revista de Educação Física.*
- Bedo BLS, Mantoan A, Catelli DS, Cruaud W, Reggiani M, Lamontagne M. 2021. BOPS: a Matlab toolbox to batch musculoskeletal data processing for OpenSim. *Computer methods in biomechanics and biomedical engineering.* 1-11.
- Bedo BLS, Pereira DR, Moraes R, Kalva-Filho CA, Will-de-Lemos T, Santiago PRP. 2020. The rapid recovery of vertical force propulsion production and postural sway after a specific fatigue protocol in female handball athletes. *Gait & posture.* 77:52-58.
- Bencke J, Curtis D, Krogshede C, Jensen LK, Bandholm T, Zebis MK. 2012. Biomechanical evaluation of the side-cutting manoeuvre associated with ACL injury in young female handball players. *Knee surgery, sports traumatology, arthroscopy : official journal of the ESSKA.* 21(8):1876-1881.
- Benjaminse A, Habu A, Sell TC, Abt JP, Fu FH, Myers JB, Lephart SM. 2008. Fatigue alters lower extremity kinematics during a single-leg stop-jump task. *Knee surgery, sports traumatology, arthroscopy : official journal of the ESSKA.* 16(4):400-407.
- Benjaminse A, Webster KE, Kimp A, Meijer M, Gokeler A. 2019. Revised Approach to the Role of Fatigue in Anterior Cruciate Ligament Injury Prevention: A Systematic Review with Meta-Analyses. *Sports medicine (Auckland, NZ).* 49(4):565-586.
- Bergmann G, Graichen F, Rohlmann A. 1993. Hip joint loading during walking and running, measured in two patients. *J Biomech.* 26(8):969-990.
- Boden BP, Dean GS, Feagin JA, Garrett WE. 2000. Mechanisms of anterior cruciate ligament injury. *Orthopedics.* 23(6):573-578.

Borotikar BS, Newcomer R, Koppes R, McLean SG. 2008. Combined effects of fatigue and decision making on female lower limb landing postures: central and peripheral contributions to ACL injury risk. *Clinical biomechanics* (Bristol, Avon). 23(1):81-92.

Bourne MN, Webster KE, Hewett TE. 2019. Is Fatigue a Risk Factor for Anterior Cruciate Ligament Rupture? *Sports medicine* (Auckland, NZ). 49(11):1629-1635.

Catelli DS, Wesseling M, Jonkers I, Lamontagne M. 2019. A musculoskeletal model customized for squatting task. 22(1):21-24.

Effects of cut angle and online processing on cutting maneuvers. *American Society of Biomechanics Annual Conference*; 2011.

Cesar GM, Pereira VS, Santiago PR, Benze BG, da Costa PH, Amorim CF, Serrao FV. 2011. Variations in dynamic knee valgus and gluteus medius onset timing in non-athletic females related to hormonal changes during the menstrual cycle. *The Knee*. 18(4):224-230.

Chappell JD, Herman DC, Knight BS, Kirkendall DT, Garrett WE, Yu B. 2005. Effect of fatigue on knee kinetics and kinematics in stop-jump tasks. *Am J Sports Med*. 33(7):1022-1029.

Chappell JD, Yu B, Kirkendall DT, Garrett WE. 2002. A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks. *Am J Sports Med*. 30(2):261-267.

Delp SL, Anderson FC, Arnold AS, Loan P, Habib A, John CT, Guendelman E, Thelen DG. 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE transactions on bio-medical engineering*. 54(11):1940-1950.

Dempsey AR, Lloyd DG, Elliott BC, Steele JR, Munro BJ. 2009. Changing Sidestep Cutting Technique Reduces Knee Valgus Loading. *Am J Sport Med*. 37(11):2194-2200.

Doorenbosch CA, Welter TG, van Ingen Schenau GJ. 1997. Intermuscular co-ordination during fast contact control leg tasks in man. *Brain Res*. 751(2):239-246.

Dürselen L, Claes L, Kiefer H. 1995. The influence of muscle forces and external loads on cruciate ligament strain. *Am J Sports Med*. 23(1):129-136.

Ebenbichler G, Kollmitzer J, Quittan M, Uhl F, Kirtley C, Fialka V. 1998. EMG fatigue patterns accompanying isometric fatiguing knee-extensions are different in mono- and bi-articular muscles. *Electroencephalography and clinical neurophysiology*. 109(3):256-262.

Fitts RH. 1994. Cellular mechanisms of muscle fatigue. *Physiological reviews*. 74(1):49-94.

Foster C, Florhaug JA, Franklin J, Gottschall L, Hrovatin LA, Parker S, Doleshal P, Dodge C. 2001. A new approach to monitoring exercise training. *Journal of strength and conditioning research*. 15(1):109-115.

Fregly BJ, Besier TF, Lloyd DG, Delp SL, Banks SA, Pandy MG, D'Lima DD. 2012. Grand challenge competition to predict in vivo knee loads. *J Orthop Res*. 30(4):503-513.

Gandevia SC. 2001. Spinal and supraspinal factors in human muscle fatigue. *Physiological reviews*. 81(4):1725-1789.

Gerus P, Sartori M, Besier TF, Fregly BJ, Delp SL, Banks SA, Pandy MG, D'Lima DD, Lloyd DG. 2013. Subject-specific knee joint geometry improves predictions of medial tibiofemoral contact forces. *J Biomech.* 46(16):2778-2786.

Givoni NJ, Pham T, Allen TJ, Proske U. 2007. The effect of quadriceps muscle fatigue on position matching at the knee. *J Physiol.* 584(Pt 1):111-119.

Griffin LY, Agel J, Albohm MJ, Arendt EA, Dick RW, Garrett WE, Garrick JG, Hewett TE, Huston L, Ireland ML et al. 2000. Noncontact anterior cruciate ligament injuries: risk factors and prevention strategies. *The Journal of the American Academy of Orthopaedic Surgeons.* 8(3):141-150.

Hamill J, Knutzen KM. 2006. *Biomechanical basis of human movement.* Lippincott Williams & Wilkins.

Hautier CA, Arsac LM, Deghdegh K, Souquet J, Belli A, Lacour JR. 2000. Influence of fatigue on EMG/force ratio and cocontraction in cycling. *Med Sci Sports Exerc.* 32(4):839-843.

Herrington L. 2011. Knee valgus angle during landing tasks in female volleyball and basketball players. *The Journal of Strength & Conditioning Research.* 25(1):262-266.

Hewett TE, Myer GD. 2011. The mechanistic connection between the trunk, hip, knee, and anterior cruciate ligament injury. *Exercise and sport sciences reviews.* 39(4):161-166.

Hewett TE, Myer GD, Ford KR. 2001. Prevention of anterior cruciate ligament injuries. *Current women's health reports.* 1(3):218-224.

Hewett TE, Torg JS, Boden BP. 2009. Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: lateral trunk and knee abduction motion are combined components of the injury mechanism. *Brit J Sport Med.* 43(6):417-422.

Hiemstra LA, Lo IK, Fowler PJ. 2001. Effect of fatigue on knee proprioception: implications for dynamic stabilization. *The Journal of orthopaedic and sports physical therapy.* 31(10):598-605.

Jackson BD, Wluka AE, Teichtahl AJ, Morris ME, Cicuttini FM. 2004. Reviewing knee osteoarthritis--a biomechanical perspective. *J Sci Med Sport.* 7(3):347-357.

Jacobs R, van Ingen Schenau GJ. 1992. Control of an external force in leg extensions in humans. *J Physiol.* 457:611-626.

Killen BA, Saxby DJ, Fortin K, Gardiner BS, Wrigley TV, Bryant AL, Lloyd DG. 2018. Individual muscle contributions to tibiofemoral compressive articular loading during walking, running and sidestepping. *J Biomech.* 80:23-31.

Koga H, Nakamae A, Shima Y, Iwasa J, Myklebust G, Engebretsen L, Bahr R, Krosshaug T. 2010. Mechanisms for noncontact anterior cruciate ligament injuries: knee joint kinematics in 10 injury situations from female team handball and basketball. *Am J Sports Med.* 38(11):2218-2225.

Kristianslund E, Faul O, Bahr R, Myklebust G, Krosshaug T. 2012. Sidestep cutting technique and knee abduction loading: implications for ACL prevention exercises. *Brit J Sport Med.* 48(9):779-783.

Kristianslund E, Faul O, Bahr R, Myklebust G, Krosshaug T. 2014. Sidestep cutting technique and knee abduction loading: implications for ACL prevention exercises. *Brit J Sport Med.* 48(9):779-783.

Kristianslund E, Krosshaug T. 2013. Comparison of drop jumps and sport-specific sidestep cutting: implications for anterior cruciate ligament injury risk screening. *Am J Sports Med.* 41(3):684-688.

Lai AKM, Arnold AS, Wakeling JM. 2017. Why are Antagonist Muscles Co-activated in My Simulation? A Musculoskeletal Model for Analysing Human Locomotor Tasks. *Annals of biomedical engineering.* 45(12):2762-2774.

Langevoort G, Myklebust G, Dvorak J, Junge A. 2007. Handball injuries during major international tournaments. *Scandinavian journal of medicine & science in sports.* 17(4):400-407.

Lattanzio PJ, Petrella RJ, Sproule JR, Fowler PJ. 1997. Effects of fatigue on knee proprioception. *Clinical journal of sport medicine : official journal of the Canadian Academy of Sport Medicine.* 7(1):22-27.

Laughlin WA, Weinhandl JT, Kernozek TW, Cobb SC, Keenan KG, O'Connor KM. 2011. The effects of single-leg landing technique on ACL loading. *J Biomech.* 44(10):1845-1851.

Lee MJC, Lloyd DG, Lay BS, Bourke PD, Alderson JA. 2016. Different visual stimuli affect body reorientation strategies during sidestepping. *Scandinavian journal of medicine & science in sports.*

Lee SY. 2015. Muscle activities of the rectus abdominis and rectus femoris and their ratio during leg raises performed by healthy adults. *Journal of physical therapy science.* 27(3):549-550.

Lerner ZF, DeMers MS, Delp SL, Browning RC. 2015. How tibiofemoral alignment and contact locations affect predictions of medial and lateral tibiofemoral contact forces. *J Biomech.* 48(4):644-650.

Liederbach M, Kremenic IJ, Orishimo KF, Pappas E, Hagins M. 2014. Comparison of landing biomechanics between male and female dancers and athletes, part 2: Influence of fatigue and implications for anterior cruciate ligament injury. *Am J Sports Med.* 42(5):1089-1095.

Lloyd DG, Besier TF. 2003. An EMG-driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo. *J Biomech.* 36(6):765-776.

Lloyd DG, Buchanan TS. 2001. Strategies of muscular support of varus and valgus isometric loads at the human knee. *J Biomech.* 34(10):1257-1267.

Ludwig O, Simon S, Piret J, Becker S. 2017. Differences in the Dominant and Non-Dominant Knee Valgus Angle in Junior Elite and Amateur Soccer Players after Unilateral Landing. 5(1).

- Mair SD, Seaber AV, Glisson RR, Garrett WE, Jr. 1996. The role of fatigue in susceptibility to acute muscle strain injury. *Am J Sports Med.* 24(2):137-143.
- Makaruk H, Czaplicki A, Sacewicz T, Sadowski J. 2014. The effects of single versus repeated plyometrics on landing biomechanics and jumping performance in men. *Biol Sport.* 31(1):9-14.
- Mantovani G, Lamontagne M. 2017. How Different Marker Sets Affect Joint Angles in Inverse Kinematics Framework. *J Biomech Eng.* 139(4):1-7.
- Martin V, Kerhervé H, Messonnier LA, Banfi JC, Geyssant A, Bonnefoy R, Féasson L, Millet GY. 2010. Central and peripheral contributions to neuromuscular fatigue induced by a 24-h treadmill run. *Journal of applied physiology (Bethesda, Md : 1985).* 108(5):1224-1233.
- McLean SG, Fellin RE, Suedekum N, Calabrese G, Passerallo A, Joy S. 2007. Impact of fatigue on gender-based high-risk landing strategies. *Med Sci Sports Exerc.* 39(3):502-514.
- McLean SG, Huang X, van den Bogert AJ. 2005. Association between lower extremity posture at contact and peak knee valgus moment during sidestepping: implications for ACL injury. *Clinical biomechanics (Bristol, Avon).* 20(8):863-870.
- McLean SG, Huang XM, Su A, van den Bogert AJ. 2004. Sagittal plane biomechanics cannot injure the ACL during sidestep cutting. *Clinical Biomechanics.* 19(8):828-838.
- Modenese L, Phillips AT. 2012. Prediction of hip contact forces and muscle activations during walking at different speeds. *Multibody System Dynamics.* 28(1):157-168.
- Mok KM, Bahr R, Krosshaug T. 2018. Reliability of lower limb biomechanics in two sport-specific sidestep cutting tasks. *Sports biomechanics.* 17(2):157-167.
- Monfort SM, Pradarelli JJ, Grooms DR, Hutchison KA, Onate JA, Chaudhari AMW. 2019. Visual-Spatial Memory Deficits Are Related to Increased Knee Valgus Angle During a Sport-Specific Sidestep Cut. *Am J Sports Med.* 47(6):1488-1495.
- Myklebust G, Maehlum S, Holm I, Bahr R. 1998. A prospective cohort study of anterior cruciate ligament injuries in elite Norwegian team handball. *Scandinavian journal of medicine & science in sports.* 8(3):149-153.
- Olsen OE, Myklebust G, Engebretsen L, Bahr R. 2004. Injury mechanisms for anterior cruciate ligament injuries in team handball a systematic video analysis. *Am J Sport Med.* 32(4):1002-1012.
- Olsen OE, Myklebust G, Engebretsen L, Bahr R. 2006. Injury pattern in youth team handball: a comparison of two prospective registration methods. *Scandinavian journal of medicine & science in sports.* 16(6):426-432.
- Palmieri-Smith RM, McLean SG, Ashton-Miller JA, Wojtys EM. 2009. Association of Quadriceps and Hamstrings Cocontraction Patterns With Knee Joint Loading. *J Athl Train.* 44(3):256-263.
- Pataky TC. 2010. Generalized n-dimensional biomechanical field analysis using statistical parametric mapping. *J Biomech.* 43(10):1976-1982.

Piry H, Fallahi A, Kordi R, Rajabi R, Rahimi M, Yosefi M. 2011. Handball injuries in elite Asian players. *World Applied Sciences Journal*. 14(10):1559-1564.

Posthumus M, Collins M, September AV, Schwellnus MP. 2011. The intrinsic risk factors for ACL ruptures: an evidence-based review. *The Physician and sportsmedicine*. 39(1):62-73.

Quatman CE, Hewett TE. 2009. The anterior cruciate ligament injury controversy: is "valgus collapse" a sex-specific mechanism? *Brit J Sport Med*. 43(5):328-335.

Saxby DJ, Modenese L, Bryant AL, Gerus P, Killen B, Fortin K, Wrigley TV, Bennell KL, Cicuttini FM, Lloyd DG. 2016. Tibiofemoral contact forces during walking, running and sidestepping. *Gait & posture*. 49:78-85.

Sigward SM, Cesar GM, Havens KL. 2015. Predictors of Frontal Plane Knee Moments During Side-Step Cutting to 45 and 110 Degrees in Men and Women: Implications for Anterior Cruciate Ligament Injury. *Clinical journal of sport medicine : official journal of the Canadian Academy of Sport Medicine*. 25(6):529-534.

Waldén M, Häggglund M, Magnusson H, Ekstrand J. 2011. Anterior cruciate ligament injury in elite football: a prospective three-cohort study. *Knee surgery, sports traumatology, arthroscopy : official journal of the ESSKA*. 19(1):11-19.

Wang LI. 2011. The lower extremity biomechanics of single- and double-leg stop-jump tasks. *Journal of sports science & medicine*. 10(1):151-156.

Weinhandl JT, Earl-Boehm JE, Ebersole KT, Huddleston WE, Armstrong BS, O'Connor KM. 2013. Anticipatory effects on anterior cruciate ligament loading during sidestep cutting. *Clinical biomechanics (Bristol, Avon)*. 28(6):655-663.

Wesseling M, Derikx LC, de Groote F, Bartels W, Meyer C, Verdonchot N, Jonkers I. 2015. Muscle optimization techniques impact the magnitude of calculated hip joint contact forces. *J Orthop Res*. 33(3):430-438.

Wojtys EM, Beaulieu ML, Ashton-Miller JA. 2016. New perspectives on ACL injury: On the role of repetitive sub-maximal knee loading in causing ACL fatigue failure. *J Orthop Res*. 34(12):2059-2068.

Xia R, Zhang X, Wang X, Sun X, Fu W. 2017. Effects of Two Fatigue Protocols on Impact Forces and Lower Extremity Kinematics during Drop Landings: Implications for Noncontact Anterior Cruciate Ligament Injury. *Journal of healthcare engineering*. 2017:5690519-5690519.

Yang D, Morris SF. 1999. Neurovascular anatomy of the rectus femoris muscle related to functioning muscle transfer. *Plastic and reconstructive surgery*. 104(1):102-106.

Yu B, Garrett WE. 2007. Mechanisms of non-contact ACL injuries. *Br J Sports Med*. 41 Suppl 1:i47-51.

Yu B, Lin C-F, Garrett WE. 2006. Lower extremity biomechanics during the landing of a stop-jump task. *Clinical Biomechanics*. 21(3):297-305.

Zahradnik D, Uchytíl J, Farana R, Jandacka D. 2014. Ground Reaction Force and Valgus Knee Loading during Landing after a Block in Female Volleyball Players. *J Hum Kinet.* 40(1):67-75.

Zech A, Steib S, Hentschke C, Eckhardt H, Pfeifer K. 2012. Effects of localized and general fatigue on static and dynamic postural control in male team handball athletes. *Journal of strength and conditioning research.* 26(4):1162-1168.

CAPÍTULO 7

DISCUSSÃO GERAL E CONSIDERAÇÕES FINAIS

O conjunto das obras apresentadas neste documento apresentaram questões relevantes e contemporâneas para a compreensão do fenômeno fadiga e sua possibilidade de influência no risco de lesões nos membros inferiores, mais especificamente na articulação do joelho. Além disso, a presente tese também buscou fornecer aos leitores ferramentas que possivelmente auxiliarão na elaboração e execução de futuros projetos de pesquisa.

Em cada uma das abordagens apresentadas, buscamos delinear um procedimento metodológico apropriado para responder questionamentos emergentes no meio acadêmico. Além disso, em alguns contextos, não buscamos apenas responder questionamentos próprios, mas principalmente indagações ainda abertas no meio acadêmico. No geral, os achados encontrados nos estudos apresentados permitiram melhor compreensão do tópico central desta tese. Ademais, também visamos elaborar e oferecer ferramentas que ajudassem outros pesquisadores.

De modo geral a presente tese teve como objetivo investigar se a fadiga, induzida por meio de um protocolo que respeitasse a especificidade de praticantes de handebol, afetaria a biomecânica da articulação do joelho mas, especificamente, qual seria o seu efeito na carga interna da articulação do joelho. Para tal feito, foi necessário a elaboração e aplicação de técnicas que permitissem a obtenção dos resultados que respondessem tal questionamento, assim como a elaboração de ferramentas visando otimizar o processo de análise. Em cada um dos estudos apresentados, apresentamos uma discussão detalhada dos resultados assim como as suas implicações no meio acadêmico e na interação dos resultados com os estudos subjacentes. Entretanto, encontramos-nos na necessidade de apresentar, de modo breve, uma discussão que abrangesse a indagação central deste documento.

No que concerne aos principais resultados encontrados, destaca-se, primeiramente, a elaboração de um protocolo de fadiga desenvolvido para atletas de handebol (Estudo 1). Ao decorrer dos anos, diferentes métodos foram utilizados para indução da fadiga em atletas de

diferentes modalidades; contudo, para o nosso conhecimento, até a elaboração da presente tese, desconhecia-se métodos elaborados especificamente para atletas desta modalidade que pudessem ser realizados em ambiente laboratorial. O protocolo desenvolvido mostrou-se eficiente para indução da fadiga que, por sua vez, afetou o controle postural de atletas de handebol feminino, assim como reduziu a capacidade de produção de força, na execução de um salto vertical máximo, por um curto período de tempo. Entretanto, foram encontrados resultados que levantam a necessidade de futuros estudos, principalmente no que concerne não só em entender como o próprio protocolo, mas também a prática real do handebol, afetaria os diferentes sistemas sensoriais. Entende-se que os sistemas sensoriais são afetados de modo diferente e, por isso, necessitam de mais ou menos tempo para recuperação. Contudo, independente da maneira que cada sistema foi afetado, as suas interações resultam em alterações afetam a biomecânica dos membros inferiores e, por isso, investigações sobre o efeito da fadiga devem ser realizadas durante esse curto período.

Diante disso, no Estudo 2, foi encontrado que a fadiga resultou em alterações cinemáticas que, possivelmente, colocariam a articulação em um maior risco de lesão. Entretanto, o estudo levou em consideração apenas variáveis cinemáticas e, devido a isso, seria errôneo afirmar que a carga articular e/ou risco de lesões aumentaria na situação de fadiga, dado que nenhum desses aspectos foi investigado no estudo. Tal precaução foi confirmada na realização do Estudo 5, na qual os principais resultados apontaram uma redução na força de reação na articulação tibiofemoral na situação de fadiga, assim como a redução da produção de força em alguns dos principais músculos da articulação do joelho. Diante disso, torna-se importante ressaltar que os nossos achados vão de encontro aos encontrados na literatura, na qual não existem evidências concretas que a fadiga aumenta o risco de lesões na articulação do joelho. Entretanto, para nosso conhecimento, esses são os primeiros estudos que investigaram

o efeito da fadiga na biomecânica do joelho de atletas de handebol utilizando um protocolo específico.

Embora cada um dos estudos apresentados possuam as suas limitações individuais, destaca-se ainda a falta de um estudo transversal observando atletas no ambiente real de prática esportiva. Apesar de tentarmos simular situações de jogo, é indispensável que futuras abordagens tentem averiguar alterações nas articulações dos membros inferiores durante situações reais de jogo.

Por fim, espero que o conjunto de ideias e propostas apresentadas no decorrer desse texto encorajem estudiosos em biomecânica a estruturarem novas abordagens que possibilitem maiores compreensões acerca dos temas abordados. Ainda, desejo fortemente que com o passar do tempo e a inserção de novos cientistas, projetos individuais e multicêntricos (nacionais e internacionais) possibilitem a melhoria da qualidade na pesquisa em biomecânica no cenário nacional e estimulem o crescimento no fomento e valorização da pesquisa em nosso país.

CAPÍTULO 9

APÊNDICES

APENDÍCE 1 – FICHA DE AVALIAÇÃO

FACULDADE DE MEDICINA DE RIBEIRAO PRETO Programa de Pós Graduação em Reabilitação e Desempenho Funcional
--

Análise Cinemática e Eletromiográfica da Articulação do Joelho em Atletas de Handebol e Universitárias

Nº: _____

Data da coleta: ____/____/____

Dados pessoais

Nome completo: _____

Data de nascimento: ____/____/____ Idade: ____ anos Sexo: (M) (F)

Profissão: _____

Peso: _____ kg Altura: _____ metros IMC: _____

Endereço: _____ Nº: _____

Complemento: _____ Bairro: _____ CEP: _____

Cidade: _____ Estado: _____

Telefone: (____) _____ Celular: (____) _____

E-mail: _____

Membro inferior dominante: (____) Direito (____) Esquerdo

Membro superior dominante: (____) Direito (____) Esquerdo

Histórico Esportivo

Modalidade praticada: _____ Posição: _____

Nível Competitivo: (Recreacional, Amador, Profissional) _____

Idade de início a prática esportiva: _____

Tempo de Treinamento: _____

Frequência semanal: _____ vezes por semana Horas/sessão: _____ horas

Treino em academia: (____) Sim (____) Não

Frequência semanal: _____ vezes por semana Horas/sessão: _____ horas

Nível de atividade física

Participa de treinamento sistematizado das seguintes modalidades:

Futebol: (✓) Sim () Não

Futsal: (✓) Sim () Não

Basquetebol: (✓) Sim () Não

Rúgbi: (✓) Sim () Não

Outras modalidades esportivas: (✓) Sim () Não

Qual: _____

NÍVEL DE ATIVIDADE FÍSICA (MCOUTYS; WYLIE; HUSTON, 1993)

Nível	Atividade
10 ()	Esporte competitivo com saltos, giros e torções.
8 ()	Esporte Recreacional com saltos, giros e torções.
6 ()	Corrida, bicicleta, natação ou ocasionalmente esporte com <u>pivoteamento</u> .
4 ()	Não realizam esporte com saltos, giros e torções, mas regularmente corrida, bicicleta ou natação.
2 ()	Não realizam esporte com saltos, giros e torções, mas ocasionalmente corrida, bicicleta ou natação.
0 ()	Inativo, sedentário.

Histórico de lesão

Lesões prévias na articulação do joelho: (✓) Sim () Não

Qual e quando? _____

Tratamento: Sim (✓) Não ().

Qual? _____

Outras Lesões: _____

Membro acometido: _____

Tempo de lesão: _____

Tratamento: Sim (✓) Não ().

Qual? _____

Cirurgias Realizadas: Sim (☒) Não (☐)

Qual e quando? _____

Atualmente, possui dor ou incômodo na articulação do joelho. Sim (☒) Não (☐).

Faz uso de anticoncepcional: (☒) Sim (☐) Não

Se a resposta for SIM - Quando começou a utilizar? _____

Se a resposta for NÃO:

Em qual fase do seu ciclo você se encontra: _____

APÊNDICE 2 – FICHA DE COLETA

FICHA DE COLETA II

Nome: _____ Código: _____

Massa: _____ Estatura: _____ Idade: _____ Data da coleta: ____/____/____

Repouso								
Tentativa			BORG			Frequência cardíaca		
Parte I – Salto vertical								
Salto	Tentativa	F _{MAX}	F _{MAX} (PC)	FPRS	FPRS (PC)			
1								
2								
3								
Parte II – Protocolo de fadiga								
1º CICLO								
Volta	Tentativa	Tempo	Frq. Card.	BORG	F _{MAX}	F _{MAX} (PC)	FPRS	FPRS (PC)
1								
2º CICLO								
Volta	Tentativa	Tempo	Frq. Card.	BORG	F _{MAX}	F _{MAX} (PC)	FPRS	FPRS (PC)
1								
2								
3º CICLO								
Volta	Tentativa	Tempo	Frq. Card.	BORG	F _{MAX}	F _{MAX} (PC)	FPRS	FPRS (PC)
1								
2								
3								
4º CICLO								
Volta	Tentativa	Tempo	Frq. Card.	BORG	F _{MAX}	F _{MAX} (PC)	FPRS	FPRS (PC)
1								
2								
3								
4								
5º CICLO								
Volta	Tentativa	Tempo	Frq. Card.	BORG	F _{MAX}	F _{MAX} (PC)	FPRS	FPRS (PC)
1								
2								
3								
4								
5								
Momento em que fadiga								
Volta	Tentativa	Tempo	Frq. Card.	BORG	F _{MAX}	F _{MAX} (PC)	FPRS	FPRS (PC)
-								

Parte III - Pós protocolo de fadiga								
Tempo	Tentativa	Tempo	Frq. Card.	BORG	F _{MAX}	F _{MAX} (PC)	FPRS	FPRS (PC)
1 min								
2 min								
3 min								
4 min								
5 min								
6 min								
7 min								
8 min								
9 min								
10 min								
Parte IV – Intensidade total do protocolo (após 30 min)								
Tempo	Tentativa	Tempo	Frq. Card.	BORG	F _{MAX}	F _{MAX} (PC)	FPRS	FPRS (PC)
-	-	-	-		-	-	-	-

APÊNDICE 3 – APROVAÇÃO COMITÊ DE ÉTICA

PARECER CONSUBSTANCIADO DO CEP

DADOS DO PROJETO DE PESQUISA

Título da Pesquisa: Efeitos de um programa preventivo de lesões no controle postural e biomecânica da articulação do joelho

Pesquisador: Paulo Roberto Pereira Santiago

Área Temática:

Versão: 2

CAAE: 70103817.0.0000.5659

Instituição Proponente: UNIVERSIDADE DE SAO PAULO

Patrocinador Principal: Financiamento Próprio

DADOS DO PARECER

Número do Parecer: 2.249.659

Apresentação do Projeto:

Ver parecer anterior

Objetivo da Pesquisa:

Ver parecer anterior

Avaliação dos Riscos e Benefícios:

Ver parecer anterior

Comentários e Considerações sobre a Pesquisa:

Ver parecer anterior

Considerações sobre os Termos de apresentação obrigatória:

Pendências sanadas: os documentos estão adequados, apresentando todas as informações necessárias.

Conclusões ou Pendências e Lista de Inadequações:

Pendências sanadas

Considerações Finais a critério do CEP:

O projeto encontra-se APROVADO para execução. Pedimos atenção aos seguintes itens:

Endereço: Avenida Bandeirantes, 3900

Bairro: VILA MONTE ALEGRE

CEP: 14.040-907

UF: SP

Município: RIBEIRAO PRETO

Telefone: (16)3315-0494

E-mail: cep90@usp.br

USP - ESCOLA DE EDUCAÇÃO
FÍSICA E ESPORTE DE
RIBEIRÃO PRETO DA USP

Continuação do Parecer: 2.249.859

- 1) De acordo com a Resolução CNS n.º 466/2012, o pesquisador deverá apresentar relatórios semestrais (parciais e final, em função da duração da pesquisa);
- 2) Eventuais emendas (modificações) ao protocolo devem ser apresentadas, com justificativa, ao CEP de forma clara e sucinta, identificando a parte do protocolo a ser modificada. Neste caso, o pesquisador deve aguardar nova aprovação do CEP para realizar os procedimentos de acordo com as mudanças solicitadas;
- 3) Sobre o TCLE: caso o termo tenha DUAS páginas ou mais, lembramos que no momento da sua assinatura, tanto o participante da pesquisa (ou seu representante legal) quanto o pesquisador responsável deverão RUBRICAR todas as folhas, colocando as assinaturas na última página;
- 4) Caso haja instituição(ões) coparticipante(s) no projeto, atender a solicitação da carta n.º 0212/CONEP/CNS, de 21 de outubro de 2010;
- 5) GARANTIR QUE NOS MOMENTOS DE ATIVIDADE FÍSICA SEMPRE HAJA PELO MENOS UM MEMBRO DA EQUIPE DE PESQUISA APTO A PRESTAR OS SOCORROS DE URGÊNCIA, INCLUSIVE MASSAGEM CARDÍACA E USO DO DEA, CASO NECESSÁRIO.

Este parecer foi elaborado baseado nos documentos abaixo relacionados:

Tipo Documento	Arquivo	Postagem	Autor	Situação
Informações Básicas do Projeto	PB_INFORMAÇÕES_BÁSICAS_DO_PROJETO_936022.pdf	20/07/2017 16:53:28		Aceito
TCLE / Termos de Assentimento / Justificativa de Ausência	TCLE_GT_v2.pdf	20/07/2017 16:53:06	Paulo Roberto Pereira Santiago	Aceito
TCLE / Termos de Assentimento / Justificativa de Ausência	TCLE_GC_v2.pdf	20/07/2017 16:52:48	Paulo Roberto Pereira Santiago	Aceito
Projeto Detalhado / Brochura Investigador	ProjetoDoutorado_BrunoBedo.pdf	21/06/2017 18:07:11	Paulo Roberto Pereira Santiago	Aceito
Folha de Rosto	FolhaDeRosto_BrunoBedo_Assinada.	06/06/2017	Paulo Roberto	Aceito

Endereço: Avenida Bandeirantes, 3900

Bairro: VILA MONTE ALEGRE

CEP: 14.040-907

UF: SP

Município: RIBEIRÃO PRETO

Telefone: (16)3315-0494

E-mail: cep90@usp.br

Página 02 de 03

APÊNDICE 4 – TERMO DE CONSENTIMENTO LIVRE E ESCLARECIDO

UNIVERSIDADE DE SÃO PAULO
ESCOLA DE EDUCAÇÃO FÍSICA E ESPORTE DE RIBEIRÃO
PRETO

Termo de Consentimento Livre e Esclarecido **(Grupo Treinamento Convencional)**

Pesquisador Responsável: Prof. Dr. Paulo Roberto Pereira Santiago

Prezado(a) Senhor(a),

Você está sendo convidado a participar como voluntário(a) do projeto de pesquisa intitulado “**Efeitos de um programa preventivo de lesões no controle postural e biomecânica da articulação do joelho antes e após um protocolo de fadiga**” a ser realizado com atletas.

A pesquisa objetiva identificar avaliar o efeito de um treinamento preventivo de lesões na articulação do joelho e no controle postural em atletas de handebol com e sem os efeitos da fadiga. Esperamos, ao término do estudo, contribuir com informações que possibilitem discussões sobre a importância de treinamentos preventivos para a redução da ocorrência de lesões no joelho de praticantes de handebol, visando melhorar o desempenho dos atletas.

Procedimentos

Caso aceite o convite, você participará de duas visitas ao Laboratório de Biomecânica e Controle Motor (LaBioCoM) na Escola de Educação Física e Esporte de Ribeirão Preto (EEFERP). Será realizada uma entrevista onde você responderá um questionário contendo informações sobre suas condições de saúde, histórico de lesões e vivência esportiva. Em seguida, você passará por uma avaliação clínica do joelho, com os principais testes de instabilidade ligamentar do joelho: Gaveta Anterior, Lackman, Pivot Shift, Gaveta Posterior, Estresse em Valgo e Varo e, por fim, por um teste de percepção de movimento da articulação do joelho. Após esses procedimentos, serão coletadas medidas do seu peso e altura e depois serão colocadas sob sua pele bolinhas de isopor com adesivo.

Posteriormente, você participará de uma sequência de testes como permanecer na postura ereta, com os olhos vendados por 40s; Aterrissar de uma plataforma, de 40 cm, com uma perna e; Realizar uma mudança de direção de 45° após uma corrida de cinco metros. Esses procedimentos serão realizados em dois momentos, antes e após uma série de tarefas de handebol (Saltos, Passes, Arremessos e Mudanças de direção) realizadas em forma de circuito dentro do laboratório com intuito de indução a fadiga. O tempo estimado para a realização de todos os procedimentos é de aproximadamente 90 minutos. Todos os testes serão filmados e gravados para análise dos movimentos dos joelhos.

Você não será submetido(a) a nenhum procedimento sem o seu conhecimento e consentimento, e pode sair desta pesquisa a qualquer momento, sem nenhum prejuízo, se comprometendo apenas a comunicar previamente os responsáveis pela pesquisa. Além disso, você está informada de que apenas as despesas de transporte e um lanche simples, com água, bolacha e uma fruta serão custeados pelos pesquisadores.

Possíveis benefícios

Por tratar de uma única coleta e não o acompanhamento das voluntárias por um período determinado não existem benefícios diretos esperados para a participante. Entretanto, de forma indireta, sua participação contribuirá para fomentar o planejamento de ações preventivas além de oportunizar a ampliação do entendimento sobre os mecanismos da lesão no joelho em situações com e sem fadiga entre atletas.

**UNIVERSIDADE DE SÃO PAULO
ESCOLA DE EDUCAÇÃO FÍSICA E ESPORTE DE RIBEIRÃO
PRETO**

Desconforto e risco

Todas as tarefas realizadas na presente pesquisa apresentam o mínimo de desconforto ao participante (dores musculares e cansaço), sendo todos relacionados com a prática cotidiana. Entretanto, caso seja verificado ou alegado o aparecimento de alguma situação desconfortável, os testes serão imediatamente interrompidos e você será acompanhado pela equipe responsável até o desaparecimento das sensações de desconforto. Além disso, você terá garantia de indenização diante de eventuais danos decorrentes da pesquisa conforme as Leis vigentes no país.

Por fim, um examinador estará constantemente ao seu lado. Se eventualmente alguma ajuda for necessária para a execução dos testes, você poderá solicitá-lo.

Liberdade de participação e sigilo de identidade

A participação no estudo é totalmente voluntária e você pode interrompê-la a qualquer momento sem nenhum prejuízo à sua pessoa, e por obrigação, os pesquisadores responsáveis excluirão todos os seus dados.

Todos os dados pessoais e informações obtidas neste estudo serão confidenciais. Garantimos que sua identificação será mantida em segredo. Os dados do estudo poderão ser utilizados para fins didáticos e científicos, mas a identificação dos participantes será preservada, isto é, mantida em sigilo.

Após preencher e assinar o termo de consentimento, você receberá uma via do mesmo, como de direito.

Eu, _____ portador do RG nº _____
_____, residente à _____, nº _____
_____, Complemento: _____, Bairro: _____, Cidade: _____
_____, aceito participar, voluntariamente, na presente pesquisa sob a responsabilidade
do Prof. Dr. Paulo Roberto Pereira Santiago.

Ribeirão Preto, _____ de _____ de _____

Nome por extenso

Assinatura

Prof. Dr. Paulo Roberto Pereira Santiago
Pesquisador Responsável

Se tiver dúvidas relacionadas a este estudo, contate:

Prof. Dr. Paulo Roberto Pereira Santiago
Escola de Educação Física e Esporte de Ribeirão Preto – Universidade de São Paulo – USP
Fone: (16) 3315-0346 / E-mail: paulosantiago@usp.br
Comitê de Ética em Pesquisa da Escola de Educação Física e Esporte de Ribeirão Preto - USP
Avenida Bandeirantes, 3900 - 14407-907 - Ribeirão Preto - SP - Brasil
Fone: (16) 3315-0494 / E-mail: cep90@usp.br

USP - ESCOLA DE EDUCAÇÃO
FÍSICA E ESPORTE DE
RIBEIRÃO PRETO DA USP

Continuação do Parecer: 2.249.659

Folha de Rosto	pdf	21:34:08	Pereira Santiago	Aceito
----------------	-----	----------	------------------	--------

Situação do Parecer:

Aprovado

Necessita Apreciação da CONEP:

Não

RIBEIRAO PRETO, 30 de Agosto de 2017

Assinado por:
Carlos Roberto Bueno Júnior
(Coordenador)

Endereço: Avenida Bandeirantes, 3900

Bairro: VILA MONTE ALEGRE

CEP: 14.040-907

UF: SP

Município: RIBEIRAO PRETO

Telefone: (16)3315-0494

E-mail: cep90@usp.br

Página 03 de 03

CAPÍTULO 10

ANEXOS

ANEXO 1 – SCRIPT PARA QUANTIFICAÇÃO DOS DADOS DE FORÇA

```
1 function PlatForc Bertec FPRS BrunoBedePhD
2 % *****
3 % File: PlatForc Bertec FPRS BrunoBedePhD.m
4 % Author: Bruno Luiz de Souza Bedo (c) 2014, All Rights Reserved
5 %         bruno.bedo@usp.br
6 %         Laboratório de Biomecânica e Controle Motor
7 %         Escola de Educação Física e Esporte de Ribeirão Preto
8 %         Universidade de São Paulo
9 %
10 %         Tenysson Will de Lemos (c) 2014, All Rights Reserved
11 %         tenysson@fmrp.usp.br
12 %         NAP - DCD
13 %         Av. Bandeirantes, 3900
14 %         Dept. Biomecanica, Medicina e Reabilitação do Aparelho Locomotor
15 %         Faculdade Medicina de Ribeirao Preto - Universidade de Sao Paulo
16 %
17 % Date: 06-06-2018
18 % Version: 1.1
19 % Objetivo:
20 %         Detectar o força máxima e a força máxima de reação no solo (salto
21 %         vertical)
22 clc
23 close all
24
25 global full_data time;
26 global contador;
27
28 % tempo de aquisicao = inputdlg('Insira o tempo de aquisicao',' ',1,'10');
29 % if(isempty(tempo de aquisicao))
30 %     error('Teste Cancelado')
31 % end
32
33 resp = inputdlg({'Nome do participante','Tentativa','Massa do participante
34 (Kg) '},' ',1,' ',' ',' ');
35
36 % Dados de Aquisicao
37 Fs = 5000; %5000
38 Npm = 500; %500
39
40 % tempo_de_acquisicao = str2double(tempo_de_acquisicao{1});
41 tempo_de_acquisicao = 10;
42 full_data_time = zeros(tempo_de_acquisicao*Fs,7);
43 contador = 1;
44
45 % Configurando
46 s = daq.createSession('ni');
47 s.addAnalogInputChannel('Dev1',0,'Voltage'); %Fx
48 s.addAnalogInputChannel('Dev1',1,'Voltage'); %Fy
49 s.addAnalogInputChannel('Dev1',2,'Voltage'); %Fz
50 s.addAnalogInputChannel('Dev1',3,'Voltage'); %Mx
51 s.addAnalogInputChannel('Dev1',4,'Voltage'); %My
52 s.addAnalogInputChannel('Dev1',5,'Voltage'); %Mz
53
54 s.Rate = Fs;
55 s.DurationInSeconds = tempo_de_acquisicao;
56
57 hf2 = figure('units','normalized','outerposition',[0 0 1 1]);title('Coletando força')
58 lh = addlistener(s,'DataAvailable', @(src,event) callbackplot(src,event,hf2,Npm));
59
60 s.NotifyWhenDataAvailableExceeds = Npm;
61 s.startBackground();
62
63 s.wait()
64
65 delete(lh)
66
67 calibration_matrix = diag([1000 1000 2000 600 400 300],0);
68 full_data_time(:,2:7) = full_data_time(:,2:7)*calibration_matrix;
```

```

69
70 % Filtering data
71 n = 4;
72 Wn=50/(5000/2); %cut-off = 50 Hz
73 [b,a]=butter(n,Wn);
74 dataF = filtfilt(b,a,full_data_time);
75
76 % Calculating the resulting force
77 mag_vector = sqrt(sum(dataF(:,2:4).^2,2));
78
79 % Detectando meio do sinal
80 PontCont = find(abs(diff(sign(mag_vector-20)))==2,4);
81 PontMed = mean (PontCont(2:3,1)); %Ponto médio entre o salto e aterrissagem
82
83 % Separando vetores
84 vet1 = mag_vector(1:PontMed); %Separando primeiro salto
85 vet2 = mag_vector(PontMed:end); %Separando aterrissagem
86
87 % Caculting peaks
88 [vpic1,ppic1] = findpeaks (vet1,dataF(1:PontMed,1),'SortStr','descend'); %Primeiro pico
89 (vet1(1))
89 [vpic2,ppic2] = findpeaks (vet2,dataF(PontMed:end,1),'SortStr','descend'); %Segundo
90 pico (vet2(1))
91
92 % Calculando pelo peso corporal
93 Massa = str2num(resp{3});
94 VectNorm = mag_vector./(Massa*9.81);
95
96 % Results
97 FmaxN = num2str(vpic1(1,1));
98 FmaxPc = num2str((vpic1(1,1))/(Massa*9.81));
99
100 FPRSN = num2str(vpic2(1,1));
101 FPRSPC = num2str((vpic2(1,1))/(Massa*9.81));
102
103 % Ploting FPRS
104 clf
105 fl = figure (1);
106 title('Resultante (Força pico de reação no solo)')
107 plot (dataF(:,1),mag_vector,'k');xlabel('Tempo(s)');ylabel('Força (N)')
108 hold on
109 plot(ppic1(1,1),vpic1(1,1),'ro','LineWidth',2);
110 plot(ppic2(1,1),vpic2(1,1),'bo','LineWidth',2);
111 title(['Força de reação no solo';['Fmax = ',num2str(FmaxN),' N | ',num2str(FmaxPc),'
112 Body Wight']...
113 ;[' FPRS = ',num2str(FPRSN),' N | ',num2str(FPRSPC),' Body Wight']]);
114 legend('Força resultante de reação no solo','Força Máxima aplicada no salto
115 (Fmax)','Força pico de reação no solo (FPRS)')
116
117 % Box de resultados
118 myicon = imread ('Imag Ok.jpg');
119 BoxResults = msgbox(['Fmax (Newtons) = ',num2str(FmaxN)],['Fmax (PC) = ',
120 num2str(FmaxPc)],...
121 ['FPRS (Newtons) = ',num2str(FPRSN)],['FPRS (PC) = ',
122 num2str(FPRSPC)]...
123 , 'Results - Salto vertical','custom',myicon);
124
125 % pause
126 % delete(BoxResults)
127
128 % Salving data
129 results = {resp{1} resp{2} resp{3} [tempo_de_acquisicao] [Fs] [full_data_time
130 mag_vector VectNorm] FmaxN FmaxPc FPRSN FPRSPC};
131 save(['Resultados_' resp{1} '_' resp{2} '.mat'],'results')
132
133 % save([cell2mat(resp(1)),'_',cell2mat(resp(2)),'.csv'],'results','-ascii')
134 end
135
136 function callbackplot(~,event,hf2,np)

```

```

131
132 global full_data time;
133 global contador;
134
135
136 full_data_time((contador-1)*np+1:np*contador,1) = event.TimeStamps;
137 full_data_time((contador-1)*np+1:np*contador,2:7) = event.Data;
138
139 figure(hf2)
140 plot(full_data_time(1:np*contador,1),2000*full_data_time(1:np*contador,4))
141 xlabel('tempo (s)')
142 ylabel('Peso (N)')
143 %title(round(20000*max(event.Data))/10)
144
145 contador = contador + 1;
146 end

```

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

The rapid recovery of vertical force propulsion production and postural sway after a specific fatigue protocol in female handball athletes

Bruno Luiz Souza Bedo^{a,b,*}, Dayanne Rodrigues Pereira^{a,c,1}, Renato Moraes^{a,b,2}, Carlos Augusto Kalva-Filho^{b,2}, Tenyson Will-de-Lemos^{a,1}, Paulo Roberto Pereira Santiago^{a,b,2}

^a Biomechanics and Motor Control Laboratory, School of Physical Education and Sport of Ribeirão Preto, University of São Paulo, Ribeirão Preto, São Paulo, Brazil

^b Rehabilitation and Functional Performance Graduate Program, Ribeirão Preto Medical School, University of São Paulo, Ribeirão Preto, São Paulo, Brazil

^c Ribeirão Preto Medical School, University of São Paulo, Ribeirão Preto, São Paulo, Brazil

ARTICLE INFO

Keywords:

Fatigue
Postural sway
Force
Handball

ABSTRACT

Background: Muscle fatigue is characterized by a decrease in muscle performance, accompanied by an increase in perceived exertion to produce a desired amount of force. The duration of the fatigue effects is not completely clear regarding postural sway and force production in athletes.

Research question: What is the effect of a specific fatigue protocol in postural sway and force production in female handball athletes and the recovery time necessary for these variables?

Methods: Twenty female handball athletes participated in this study. Specific handball actions composed the fatigue protocol in the format of a circuit with the gradual increment of laps. They stood upright in one-leg posture on a force plate. For the maximum propulsion force (F_{MAX}), they performed a countermovement jump on the force plate. The center of pressure (COP) and the F_{MAX} were obtained at baseline, immediately after the exhaustion due to the fatigue protocol and every minute during the first 10 min of the recovery phase.

Results: The F_{MAX} decreased ~9.5% after the fatigue protocol, whereas the COP area increased 224%. During the recovery phase, the F_{MAX} and the COP area returned to baseline values after five minutes.

Significance: The first five minutes immediately after the fatigue protocol represent the time interval where the fatigue effects are still present. Future studies willing to investigate the effects of fatigue in athletes should perform the analysis within this short time-window.

1. Introduction

Muscle fatigue results from a complex interaction between different body systems and is defined as an exercise-induced reduction in the ability to exert muscle force or power [1]. Among several deleterious effects for movement performance, fatigue relates to modification on body homeostasis, especially in the proprioceptive system [2,3], which in turn contributes to decreasing joint stability and compromises the ligaments' stability [3,4]. In addition, postural control is impaired [4,5]. These deleterious effects have been observed in athletes of different modalities [6–9].

However, the effects of fatigue on balancing tasks are not entirely understood [9], as well as the time of recovery of the postural sway and

force parameters. Fox et al. [10] observed the adverse effects of fatigue on postural stability until the thirteenth minute using either aerobic or anaerobic exercises. Differently, Guidetti et al. [11] found that the impact of fatigue remained until the tenth minute after exercising at the anaerobic threshold and until the fifth minute after exercising at the post-ventilatory threshold. Finally, Dickin and Doan [12] observed that the recovery of the postural control system remained up to 30 min after the fatigue protocol. Although the influence of fatigue on the postural control is quite clear, results are divergent about the magnitude of the adverse effects and the time-course of the recovery phase. The protocol used to induce fatigue must consider the specificity of the sport modality, as the time-course of the recovery phase is specific to the fatigue protocol used.

* Corresponding author at: Biomechanics and Motor Control Laboratory, School of Physical Education and Sport of Ribeirão Preto, University of São Paulo, Av. Bandeirantes, 3900, Monte Alegre, 14040-907, Ribeirão Preto, SP, Brazil.

E-mail addresses: bruno.bedo@usp.br (B.L.S. Bedo), daydrodriguesprra@gmail.com (D.R. Pereira), renatomoraes@usp.br (R. Moraes), kalvafilho@yahoo.com.br (C.A. Kalva-Filho), tenyson@fmrp.usp.br (T. Will-de-Lemos), paulosantiago@usp.br (P.R.P. Santiago).

¹ Address: Av. Bandeirantes, 3900, Monte Alegre, 14040-900, Ribeirão Preto, São Paulo, Brazil.

² Address: Av. Bandeirantes, 3900, Monte Alegre, 14040-907, Ribeirão Preto, São Paulo, Brazil.

<https://doi.org/10.1016/j.gaitpost.2020.01.017>

Received 19 July 2019; Received in revised form 11 December 2019; Accepted 20 January 2020

0966-6362/ © 2020 Elsevier B.V. All rights reserved.

Journal of Sports Sciences

The effect of fatigue on hip and knee kinematics during single- and double-leg dynamic tasks: A study on female handball players
--Manuscript Draft--

Full Title:	The effect of fatigue on hip and knee kinematics during single- and double-leg dynamic tasks: A study on female handball players
Manuscript Number:	
Article Type:	Original Manuscript
Keywords:	Handball; Fatigue; Knee; Kinematics; sidestep cutting maneuver
Abstract:	<p>This study investigated the effect of a handball-specific fatigue protocol in the hip and knee kinematics. Twenty female handball athletes performed three trials of the single-leg landing(SLL), sidestep cutting maneuver(SCM), and drop vertical jump(DVJ) before and after the fatigue protocol. Knee and hip angles waveforms were compared using the statistical parametric mapping, with $p < 0.05$. During the SLL, the fatigue increased hip adduction (4-7% cycle, $p = 0.045$) and knee abduction (4-9%, $p = 0.037$; 25-27%, $p = 0.048$). For the SCM, the fatigue reduced the hip flexion during 14-29% and 44-68% of the cycle ($p = 0.031$ and 0.014, respectively). Similarly, the knee flexion decreased between 7-36% and 53-73% ($p = 0.0001$ and 0.0017, respectively). Besides, during the fatigue state, the athletes reduced the hip abduction between 0-11% of the cycle ($p = 0.043$) and increased the knee abduction between 20-23% ($p = 0.043$). During the DVJ task, the fatigue decreased the hip flexion between 19-44% of the cycle ($p = 0.002$) and the knee flexion between 1-16% and 18-77%. The fatigue protocol altered the lower limb kinematics, decreasing the knee and hip flexion during the SCM and DVJ, and increasing the knee valgus during both single-leg landing tasks.</p>
Order of Authors:	Bruno Luiz de Souza Bedo, MD Danilo S. Catelli, PhD Renato Moraes, PhD Mario Lamontagne, PhD Dayanne Rodrigues Pereira Julia Barachini Graça Paulo Roberto Pereira Santiago, PhD

A custom musculoskeletal model for estimation of medial and lateral tibiofemoral contact forces during tasks with high knee and hip flexions

Bruno Luiz Souza Bedo^{a,b,c}, Danilo S. Catelli^b, Mario Lamontagne^b and Paulo Roberto Pereira Santiago^{a,c}

^aRibeirão Preto Medical School, University of São Paulo, Ribeirão Preto, SP, Brazil; ^bSchool of Human Kinetics, University of Ottawa, Ottawa, ON, Canada; ^cSchool of Physical Education and Sports of Ribeirão Preto, University of São Paulo, Ribeirão Preto, SP, Brazil

ABSTRACT

Most of musculoskeletal models (MSKM) estimate the tibiofemoral joint reaction load at a single point or do not support large lower-limb ranges. This study aimed to adapt a generic MSKM that allows large knee and hip flexions to compute medial and lateral tibiofemoral contact forces (TFCF) during gait and squat tasks. The updated model includes medial and lateral knee compartment geometries that allow computing the vertical TFCF. The updated MSKM does not affect kinematics and kinetics outputs in both of the tasks, and the sum of the medial and lateral TFCF was equivalent to the net TFCF of the original MSKM.

ARTICLE HISTORY

Received 17 January 2020
Accepted 15 April 2020

KEYWORDS

Musculoskeletal model;
tibiofemoral; knee; knee
contact forces; squat; gait

Introduction

The inverse dynamic approach has been used to indirectly measure the load on the medial and lateral condyles (Lloyd and Besier 2003); however, this approach still does not take account for muscle forces. Tibiofemoral alignment asymmetry affects the load redistribution on the medial and lateral components, increasing the risks of osteoarthritis (Jackson et al. 2004). Therefore, estimation of knee contact loading in the medial and lateral compartments would provide a beneficial assessment (Manal and Buchanan 2013).

Musculoskeletal models (MSKM) have been efficient to investigate human movement and predict the effect of interventions in a non-invasive fashion (Delp et al. 2007) as well as to estimate muscle forces and joint contact loading (Lerner et al. 2015; Rajagopal et al. 2016; Catelli et al. 2019). In order to account for a better muscle force solution in a young cohort, Rajagopal et al. (2016) used magnetic resonance imaging of young healthy individuals to redefine musculo-tendon parameters (Figure 1B). This MSKM model was further revised twice when increasing the knee range and updating muscle-tendon units (MTU) force-generating properties and paths (Figure 1C) (Lai et al. 2017), and by controlling the moment arm of hip and knee muscles to enable investigations with high range of motion (Figure 1D) (Catelli et al. 2019).

Still, these MSKMs calculate the knee contact load at a single point in the tibiofemoral joint. Resolving the magnitudes of medial and lateral forces by approximating compartment points of contact is an approach for estimating contact forces (Winby et al. 2009; Kumar et al. 2012; Gerus et al. 2013), which has been already implemented (Lerner et al. 2015) in a preceding MSKM (Delp et al. 1990), that still lacks muscle paths reliability during high lower-limb flexions and has its musculotendon parameters based on cadaveric data (Figure 1A).

Therefore, the purpose of this study was to examine the effect of including the method that computes medial and lateral TFCF to an MSKM that allows large lower-limb joint ranges and test its reliability during gait and deep squatting tasks. We hypothesized that the sum of the medial and lateral vertical components of the TFCF of the updated MSKM should match with the single vertical TFCF of the original MSKM during both tasks, as well as the joint kinematics and kinetics outcomes should not differ between the two models.

Methods

Generic model

The generic MSKM used in this study has 80 lower-limb Hill-type MTU, 37 degrees of freedom, and 17

CONTACT Paulo Roberto Pereira Santiagopaulosantiago@usp.br Biomechanics Motor Control Laboratory School of Physical Education and Sport of Ribeirão Preto, University of São Paulo, Av. Bandeirantes 3900, Ribeirão Preto, SP, 14040-907, Brazil.

© 2020 Informa UK Limited, trading as Taylor & Francis Group

BOPS: a Matlab toolbox to batch musculoskeletal data processing for OpenSim

Bruno L. S. Bedo^{a,b} , Alice Mantoan^c , Danilo S. Catelli^b , Willian Cruaud^b , Monica Reggiani^c
and Mario Lamontagne^b

^aRibeirao Preto Medical School, University of Sao Paulo, Ribeirao Preto, SP, Brazil; ^bSchool of Human Kinetics, University of Ottawa, Ottawa, ON, Canada; ^cDepartment of Management and Engineering, University of Padova, Vicenza, Italy

ABSTRACT

This paper presents Batch OpenSim Processing Scripts (BOPS), a Matlab toolbox for batch processing common OpenSim procedures: Inverse Kinematics, Inverse Dynamics, Muscle Analysis, Static Optimization, and Joint Reaction Analysis. BOPS is an easy-to-use and highly configurable tool that aims to reduce the time required to process large datasets, thus fostering the adoption of musculoskeletal modeling and simulations in daily practice. Its graphical user interface includes pre-defined setup files and has been designed to fulfill the needs of different research projects by simplifying the customization of the procedures, facilitating the analysis, and boosting research group collaborations. BOPS is released under Apache License 2.0, and its source code is freely available on SimTK and GitHub.

ARTICLE HISTORY

Received 18 June 2020
Accepted 20 December 2020

KEYWORDS

Musculoskeletal simulations;
OpenSim; data processing;
Matlab toolbox

Introduction

Musculoskeletal modeling provides a non-invasive approach to estimate muscle forces and joint contact loading, and it is an essential tool to better understand the musculoskeletal system's actions during human movement (Delp et al. 2007; Seth et al. 2018). However, it usually requires to collect motion data and to process and export them to obtain input data for the simulations. Besides, applications may require personalized musculoskeletal models, which thus must be generated. Commercial solutions and free tools are available to the biomechanics community to simplify each step within the workflow of neuro- and musculoskeletal modeling (Figure 1). For instance, BTK (Barre and Armand 2014) allows the reading and writing of C3D files, the standard file format in the field (C3D—Coordinate 3D, Motion Lab Systems), while Mokka, a standalone application by the same authors, was conceived for their visual inspection; the MoCap (Burger and Toiviainen 2013) and biomechZoo (Dixon et al. 2017) are two toolboxes enabling motion data analysis and visualization; ADAT toolbox (James and Wixted 2011) is intended to handle sports movement data; pyomeca (Martinez et al. 2020) enables data extraction, processing and

visualization for use in research and education. MOtoNMS (Mantoan et al. 2015) processes experimental data from different motion capture devices and generates inputs for neuromusculoskeletal modeling, while CEINMS (Pizzolato et al. 2015) is the first open-source toolbox that allows individual neural drive from EMG to be used as input into simulations. NMS Builder (Valente et al. 2017) was conceived as a free software application to generate subject-specific musculoskeletal models for OpenSim (Stanford University, Stanford, USA), as well as the open-source Musculoskeletal Atlas Project (MAP) Client software (Zhang et al. 2014), that provides workflows to generate musculoskeletal models from clinical data using statistical models.

Most of the listed tools have been developed in Matlab (Mathworks Inc., Natick, USA), indeed only those requiring high computational performances (CEINMS) or conceived as a standalone application (i.e. NMS Builder and OpenSim) are based on C++ programming language (Table 1). Therefore, Matlab turns out to be the preferred development tool for data processing and analysis in biomechanics, likely due to its general availability within research groups and the basic programming skills required.

CONTACT Monica Reggiani monica.reggiani@unipd.it
B.L.S.B., A.M. and D.S.C. contributed equally as first authors.
© 2021 Informa UK Limited, trading as Taylor & Francis Group