Middlesex University Research Repository

An open access repository of

Middlesex University research

<http://eprints.mdx.ac.uk>

Li, Junyan ORCID logoORCID: <https://orcid.org/0000-0003-4053-8334>, Marra, Marco A., Verdonshot, Nico and Lu, Yongtao (2021) A three-dimensional finite-element model of gluteus medius muscle incorporating inverse-dynamics-based optimization for simulation of non-uniform muscle contraction. Medical Engineering and Physics, 87 . 38 - 44. ISSN 1350-4533 [Article] (doi:10.1016/j.medengphy.2020.11.009)

Final accepted version (with author's formatting)

This version is available at: <https://eprints.mdx.ac.uk/31439/>

Copyright:

Middlesex University Research Repository makes the University's research available electronically.

Copyright and moral rights to this work are retained by the author and/or other copyright owners unless otherwise stated. The work is supplied on the understanding that any use for commercial gain is strictly forbidden. A copy may be downloaded for personal, non-commercial, research or study without prior permission and without charge.

Works, including theses and research projects, may not be reproduced in any format or medium, or extensive quotations taken from them, or their content changed in any way, without first obtaining permission in writing from the copyright holder(s). They may not be sold or exploited commercially in any format or medium without the prior written permission of the copyright holder(s).

Full bibliographic details must be given when referring to, or quoting from full items including the author's name, the title of the work, publication details where relevant (place, publisher, date), pagination, and for theses or dissertations the awarding institution, the degree type awarded, and the date of the award.

If you believe that any material held in the repository infringes copyright law, please contact the Repository Team at Middlesex University via the following email address:

eprints@mdx.ac.uk

The item will be removed from the repository while any claim is being investigated.

See also repository copyright: re-use policy: <http://eprints.mdx.ac.uk/policies.html#copy>

A three-dimensional finite-element model of gluteus medius muscle incorporating inverse-dynamics-based optimization for simulation of non-uniform muscle contraction

Junyan Li^{1*}, Marco A. Marra², Nico Verdonshot^{2,3}, Yongtao Lu^{4*}

Institute:

¹School of Science and Technology, Middlesex University, London, UK

²Department of Biomechanical Engineering, University of Twente, The Netherlands

³Orthopaedic Research Laboratory, Radboud Institute for Health Sciences, Radboud University Medical Center, The Netherlands

⁴Department of Engineering Mechanics, Dalian University of Technology, Dalian, China

Key words: 3D finite element muscle model; non-uniform muscle contraction; gluteus medius; optimization of muscle contraction; musculoskeletal modelling

Corresponding author:

Junyan Li

lijerry@gmail.com

School of Science and Technology, Middlesex University, London, United Kingdom

Yongtao Lu

yongtaolu@dlut.edu.cn

Department of Engineering Mechanics, Dalian University of Technology, Dalian, China

1 **Abstract**

2 Non-uniform contraction exists in many skeletal muscles and plays an important role in the
3 function of the musculoskeletal system. Particularly in the gluteus medius (GM) muscle, its three
4 subdivisions appear activated differently while performing various motion tasks. However, the non-
5 uniform contractile mechanism of GM is poorly understood. In this study, a three-dimensional finite
6 element (FE) model of GM was developed. Non-uniform contraction patterns of the three
7 subdivisions of GM during abduction, internal and external rotation were simulated through an
8 inverse-dynamics-based optimization approach. A set of sensitivity studies were also undertaken to
9 evaluate the influence of parameters including the cost function of optimization and dimension of
10 GM subdivisions on the predicted non-uniform contraction and biomechanics of the muscle.
11 Contraction across GM was found to be highly non-uniform during various motions. The whole GM
12 was activated during abduction, whereas only the anterior and posterior subdivisions were primarily
13 involved in internal and external rotation, respectively. The active contractile stress in a subdivision
14 during abduction was increased if its proportion in GM was expanded. The cost functions of
15 minimizing the sum of active contractile stresses squared/cubed provide similar qualitative
16 predictions of the trend of results. This approach provides the methodological basis to enable
17 simulation of non-uniform muscle contraction using 3D musculoskeletal models.

19

20

21 **1. Introduction**

22 Muscles play an important role in the function of the musculoskeletal system. Dysfunction of
23 muscles leads to aberrant postures, non-physiological loads on joints and ligaments and, eventually,
24 the progression of musculoskeletal disorders [1, 2]. The gluteus medius (GM) muscle is the prime hip
25 abductor. Dysfunction of GM has been implicated in impaired gait patterns, low back pain,
26 patellofemoral pain syndrome and many other lower limb injuries [3, 4].

27 A muscle usually consists of a number of motor neurons. When a motor neuron is activated,
28 the muscle fibres innervated by the motor neuron become stimulated and contract. The electro-
29 mechanics behaviour of muscles is closely associated with the spatio-temporal pattern of muscles [5-
30 7]. As a motor neuron can be activated individually, contraction in many skeletal muscles, i.e. the
31 contractile activity indicated by active contractile forces/stresses, is non-uniform [8, 9]. Particularly
32 in GM, there are three anatomically distinct subdivisions with potential for independent neural control,
33 and as a result, the three subdivisions of GM appear activated differently while performing various
34 motion tasks [8, 10]. The anterior and medial subdivisions that have fibres oriented more parallel to
35 the femur would be better positioned to abduct the hip than the posterior portion. The tilt of fibres in
36 the anterior and posterior subdivisions with respect to the frontal plane suggests their roles in hip
37 internal and external rotation, respectively [10, 11].

38 There are a number of studies on non-uniform muscle contraction, mostly using
39 electromyography (EMG) [8, 11-13] or FDG-PET (monitor fluorodeoxyglucose (FDG) uptake
40 through positron emission tomography (PET)) [14]. However, the reliability of using EMG as an
41 approach to measure biomechanics of muscles is questionable due to uncertainties in the acquisition
42 and conversion of EMG signals into biomechanical response of muscles. Apart from the substantial
43 artefacts introduced by cross-talk from neighbouring muscles or from adjacent subdivisions of the
44 same muscle [15], the use of surface EMG is inappropriate for the whole GM because the posterior
45 GM is completely covered by the gluteus maximus muscle [8]. On the other hand, intramuscular
46 EMG or FDG-PET is invasive and would alter the normal function of muscles [16, 17]. Furthermore,
47 previous experimental studies do not explain the cause-and-effect relationship between non-uniform
48 muscle contraction and motion, the synergistic mechanism among muscle subdivisions, or the
49 sensitivity of muscle activities to parameters such as the dimension and strength of a subdivision.

This information is important for planning surgical treatments and rehabilitation programmes to reduce pain and disability involved in musculoskeletal disorders [18], but remains poorly understood.

Computer models serve as an alternative approach with the potential to provide comprehensive biomechanical analyses of non-uniform muscle contraction. Most of the previous computer models that offer predictive muscle contraction are based on multibody dynamics musculoskeletal simulations with muscles assumed as one-dimensional (1D) line-segment models [19, 20]. Using these musculoskeletal models, muscle forces can be calculated based on non-invasively measured kinematics. However, 1D muscle models have limitations for simulations of non-uniform muscle contraction, because such models lack the ability to incorporate realistic three-dimensional (3D) muscle geometry and spatial fibre architecture within muscles, which would limit their modelling accuracy [21]. Furthermore, 1D models do not provide parameters such as stresses, strains and distribution of non-uniform contraction of a muscle. These parameters are important for systematic biomechanical evaluation of muscles.

Muscle models with realistic 3D geometry and detailed fibre architecture have been developed using finite element (FE) method [21-24]. However, in these previous 3D muscle models, muscle contractions were among the inputs rather than being calculated, because the redundancy issue, i.e. muscles outnumber equations of equilibrium requiring optimization to determine a unique solution of muscle contractions, was not addressed. This hampers the application of previous 3D models to study non-uniform muscle contraction. Furthermore, FE models of GM is not available yet. In a recent forward dynamics-based 3D FE musculoskeletal model [25], proportional-integral-derivative (PID) controllers were introduced to calculate muscle activations. Recently, we have developed an inverse dynamics-based FE musculoskeletal model with the ability to predict 3D muscle contractions based on kinematic data [26]. However, in these models, the contraction across each muscle was assumed to be uniform.

Therefore, the aim of this study was to develop a 3D FE model of the GM muscle incorporating inverse-dynamics-based optimization that was capable of simulating non-uniform contraction of the muscle during different motions. Additionally, a set of sensitivity studies were undertaken to evaluate the influence of parameters including the cost function of optimization and dimension of GM subdivisions on the predicted non-uniform contraction and biomechanics of the muscle.

2. Materials and Methods

In this study, subject-specific 3D geometric models reconstructed from magnetic-resonance-imaging (MRI) in the TLEM 2.0 database [27] including the GM muscle, femur and pelvis of the right lower extremity were adopted. Using a meshing software package (Hypermesh V2017, Altair, USA), the muscle and bones were represented with 4605 eight-noded hexahedral elements and 23743 four-noded tetrahedral elements, respectively (**Fig. 1a**). The mesh was dense enough to ensure the change in the peak tensile stress and active contractile stress was within 5% if the number of elements of the muscle was doubled.

Through the insertion/origin sites of the muscle/tendon tissue that are based on the anatomical information provided in TLEM 2.0 [27], the muscle/tendon tissue was rigidly attached to the bones. As deformation of joints and bones was not the subject of interest in this study and was minimal comparing to the deformation of muscles [28], the hip was assumed as a three degrees-of-freedom ball-and-socket joint and bones as rigid in order to enhance computational efficiency.

Both active and passive properties of the muscles were considered. Incompressible transversely isotropic Mooney-Rivlin material [29] incorporating fibres was adopted to represent the tendons and passive properties of the muscles. The fibre orientation within GM (**Fig. 1b**) and the regions of tendon and muscle tissues (**Fig. 2**) were identified using anatomical knowledge [10, 30]. The strain energy W of this constitutive model was given in [31]:

$$W = F_1(I_1, I_2) + F_2(\lambda) + \frac{K}{2} [\ln(J)]^2 \quad (1)$$

where, the function F_1 represents the material response of the isotropic ground substance matrix in the form of Mooney-Rivlin material as described in Eq. (2); F_2 contribution from the fibres as illustrated in Eq. (3); J volume ratio; K a bulk modulus-like penalty parameter.

$$F_1 = \frac{C_1}{2} (I_1 - 3) + \frac{C_2}{2} (I_2 - 3) \quad (2)$$

where, C_1 and C_2 are the Mooney-Rivlin material coefficients; I_1 and I_2 first and second strain invariants of the deviatoric Cauchy–Green tensor C .

$$\lambda \frac{\partial F_2(\lambda)}{\partial \lambda} = \begin{cases} 0 & \lambda \leq 1 \\ C_3(e^{C_4(\lambda-1)} - 1) & 1 < \lambda < \lambda_m \\ C_5\lambda + C_6 & \lambda \geq \lambda_m \end{cases} \quad (3)$$

where, λ is the deviatoric part of the stretch along the fibre direction; λ_m stretch at which the fibres are straightened; C_3 scales the exponential stresses; C_4 rate of uncrimping of the fibres; C_5 modulus of the straightened fibres; C_6 determined to ensure continuous stress at λ_m . λ is usually

111 bigger than zero as it reflects fibre stretch; $\lambda < 1$ describes the material in compression; $\lambda > 1$
 112 describes the material in tension. Active contraction along the fibres was incorporated into the muscle
 113 material model (**Fig. 1b**), with the total stress in the solid mixture (σ) as the sum of the solid stress
 114 due to strain (σ^s) and the active contractile stress (σ^a):

$$115 \quad \sigma = \sigma^s + \sigma^a \quad (4)$$

116 Values of the parameters in the constitutive model are shown in **Table 1**. Three subdivisions
 117 were defined in the GM model, with independent active contractile stress in each of the portions
 118 (**Fig. 2**). Uniform contraction was assumed in each subdivision. The three subdivisions were of
 119 similar dimensions in the original model (**Fig. 2a**). To assess the effect of the dimension of
 120 subdivisions on the model predictions, three other models were created in which the cross-sectional
 121 area of one of the three GM subdivisions was approximately 1.5 times larger than the other two
 122 portions (**Fig. 2b, c and d**). This choice of proportion and variation is based on anatomical
 123 observations [10, 30].

124 The pelvis was immobilized in all degrees-of-freedom. Motions that GM contributes to
 125 including abduction, internal rotation and external rotation [11] were applied to the hip joint
 126 separately (through rotating the femur), starting from 0 degrees to 20 degrees and ramped over 0.5 s.
 127 Flexion, extension and adduction were not investigated, because GM plays a minor role in these
 128 motions. Except for the insertion/origin sites of the muscle/tendon tissue connecting the bones, the
 129 GM muscle is not anatomically in contact with the bones. Therefore, contact between the muscle and
 130 bones was not defined in the model to enhance computation efficiency.

131 During each motion, there are three unknown active contractile stresses to be calculated versus
 132 one moment equilibrium equation which is along the axis of rotation of the hip joint. In order to solve
 133 this redundancy issue, optimization is needed to determine a unique solution of active contractile
 134 stresses in the subdivisions. Therefore, an inverse-dynamics-based optimization approached was
 135 developed. Based on the muscle active contractile stresses and the corresponding joint moments in
 136 the FE model, active contractile stresses in the subdivisions of the FE GM model were optimized
 137 until 1) the cost function was minimized and 2) constant resisting moments of 5.0 N×m, 0.2 N×m,
 138 and 0.5 N×m along the axis of rotation of the hip joint were generated by GM during abduction,
 139 internal rotation and external rotation, respectively (e.g. adduction moment generated during
 140 abduction). These moments would ensure a proper level of activation for GM (this is equivalent to
 141 applying torques to the joint for the muscle to balance). The two cost functions that have been widely
 142 used in recent musculoskeletal models of the lower extremity including minimizing the sum of active
 143 contractile stresses squared (i.e. $\sum_{i=1}^n (\sigma^a_i)^2$) [32] and minimizing the sum of active contractile

144 stresses cubed (i.e. $\sum_{i=1}^n (\sigma^a_i)^3$) [33] were simulated to evaluate the sensitivity of the model to the
145 cost functions. Another cost function that minimizes the maximum active contractile stress (i.e.
146 $\min(\max(\sigma^a))$) was also assessed [34]. Results of non-uniform muscle contractions under different
147 cost functions were compared to a GM model with uniform muscle contraction across the whole
148 muscle. Notably, the system with uniform GM contraction is not redundant and thus optimization is
149 not needed. Optimization and analyses were conducted at 0 degrees, 5 degrees, 15 degrees and
150 20 degrees for each motion. The muscle lengths and moment arms were recalculated at each
151 optimization step.

152 FE modelling was performed in the open-source implicit FE solver FEBio (V2.6.4;
153 <http://febio.org/febio>). To enhance computational efficiency, the FE model at each quasi-static time
154 instance (e.g. at 0.5 s) in the optimization process was simulated based on the model at the previous
155 time instance (i.e. at 0.4 s) in which the optimization criteria were achieved, rather than starting from
156 the original state (i.e. at 0 s). Optimization and automation of data transfer between the FE modelling
157 and optimization procedures were achieved in MATLAB (R2017a, Mathworks, MA). The “fmincon”
158 optimization tool in MATLAB was adopted to solve the optimization problem. The simulations were
159 performed on a Windows 10 computer with 64 GB of RAM and 32 Intel E5-2699 cores at 2.2 GHz.
160 Active contractile stresses and tensile stresses (i.e. the first principal stress of σ in Eq (4)) of the GM
161 muscle were analyzed.

162

163

164 3. Results

165 Distribution of tensile stresses across the GM model during different motions is shown in **Fig. 3**.
166 Tensile stresses were concentrated around the tissue region connecting the femur, where the tissue
167 has smaller cross-sectional area than in the rest of the GM. Tensile stresses in all the subdivisions
168 were of similar levels during hip abduction. Among the three subdivisions, tensile stresses were
169 markedly higher in the anterior subdivision and posterior subdivision during internal rotation and
170 external rotation, respectively. For each motion, the maximum tensile stress was higher at 20 degrees
171 rotation than in the original configuration.

172 As shown in **Fig. 4**, active contractile stresses increased with increasing rotation angle of all the
173 motions, in particular during internal and external rotation. The whole GM were activated during
174 abduction, while only the anterior subdivision displayed contraction during internal rotation. During
175 external rotation, both the medial and posterior subdivisions were activated from 0 degrees to 5
176 degrees, while the medial subdivision became inactivated when the rotation exceeded 10 degrees.

177 During abduction, active contractile stresses in the anterior and medial subdivisions were similar and
178 higher than the posterior portion.

179 The active contractile stresses were different under different cost functions during abduction,
180 particularly in the posterior subdivision (**Fig. 5**). Results of the models with uniform contraction and
181 with the cost function of minimizing the maximum active contractile stress were identical, but
182 different from the other two models. Under the two cost functions of minimizing the sum of active
183 contractile stresses squared/cubed, the active contractile stress increased with increasing abduction
184 angle and was higher in the anterior and medial subdivisions than the posterior portion. A higher level
185 of non-uniform contraction was observed under the cost function of minimizing the sum of active
186 contractile stresses squared than minimizing the sum of active contractile stresses cubed. As shown
187 in **Fig. 6**, there was also a difference in active contractile stresses among the models with subdivisions
188 of different dimensions. The active contractile stress was higher in a larger subdivision during
189 abduction than the same subdivision of the model with subdivisions of similar dimensions.

190

191

192 **4. Discussion**

193 In this study, non-uniform contraction of the GM muscle during different motions was
194 simulated for the first time through a novel approach combining FE modelling and inverse-dynamics-
195 based optimization. It was found that contraction in GM is highly non-uniform across its subdivisions
196 during different motions. This is generally in agreement with previous EMG and FDG-PET studies
197 [8, 11-14]. There was also a marked difference in the biomechanics of GM between the models with
198 and without the consideration of non-uniform contraction. These findings demonstrate the important
199 role of non-uniform muscle contraction while performing various motion tasks as well as the great
200 need to consider non-uniform muscle contractions in musculoskeletal models.

201 Computer models have the advantage over experimental measurements in terms of elucidating
202 the synergistic mechanism among muscles. In theory, every agonistic muscle in the musculoskeletal
203 system should be activated to some extent under the cost function of minimizing the sum of active
204 contractile stresses squared or cubed [35]. In this study, it was shown that all the GM subdivisions
205 contribute to hip abduction, while only the anterior and posterior subdivisions are primarily activated
206 during hip internal and external rotation, respectively. This would suggest that the anterior and
207 posterior subdivisions act as agonists during abduction but as antagonists to each other during internal
208 (anterior subdivision as the agonist) and external (posterior subdivision as the agonist) rotation. This
209 is also reflected by the finding that contraction in the anterior (or posterior) subdivision increases with

210 increasing internal (or external) rotation angle in order to balance the increased passive tension in the
211 posterior (or anterior) portion. These results contradict a previous surface EMG measurement, where
212 it was found that the anterior subdivision displayed higher activation than the posterior portion during
213 both internal and external rotation [13], but are supported by the anatomical observation that the
214 orientation of fibres in the anterior and posterior subdivisions would reflect their participation in
215 internal and external rotation, respectively [10]. It was found that the collaborative pattern among
216 GM subdivisions does not only depends on the type of motion, but is also subject to the level of
217 external rotation. The medial subdivision took part in the initial external rotation but became an
218 antagonist to external rotation above 10 degrees.

219 The proportion in the dimension of each subdivision in GM varies between individuals [10, 30].
220 Additionally, it is unclear whether the cost functions of optimization used in previous musculoskeletal
221 models would provide contradictory results when applied to simulate non-uniform muscle contraction.
222 The parametric study was therefore undertaken as a precursor to future model validation to identify
223 the sensitivity of the model to these parameters. The findings of this study show that contraction
224 patterns of GM are dependent on the dimension of its subdivisions. The level of contraction in a
225 subdivision during abduction was increased if its proportion in GM was expanded, because of its
226 ability of sharing a higher proportion of the hip adduction moment applied [35]. Compared to
227 previous musculoskeletal models in which the GM was typically represented by three independent
228 1D line-segments [36], this study offers more systematic analyses of the distribution of activation and
229 stresses across the 3D structure, and at the same time, accounting for the shear effect between the
230 connecting subdivisions. In this study, only the results during abduction are shown to demonstrate
231 the sensitivity of the model to the cost functions and the dimension of the muscle subdivisions. This
232 is because that only the anterior and posterior subdivisions were primarily activated during internal
233 and external rotation, respectively. As a results, various cost functions and dimensions of the muscle
234 subdivisions provided similar predictions during internal and external rotation.

235 The two cost functions of optimization that have been widely used in previous musculoskeletal
236 models of the lower extremity, i.e. minimizing the sum of active contractile stresses squared/cubed,
237 were found to result in different levels of muscle contraction during abduction, but provide similar
238 trends in the relationship between contraction and rotation angle and in the comparison of
239 contractions between subdivisions. Further experimental measurements and validation studies are
240 needed to determine the optimal cost function in future modelling of 3D muscles. Under both cost
241 functions, the level of contraction increased with increasing abduction angle and was higher in the
242 anterior and medial subdivisions than the posterior portion. This is consistent with the anatomical
243 observation that the anterior and medial subdivisions have fibres oriented more parallel to the femur

244 and thus would be better positioned to abduct the hip than the posterior portion [10]. Reliable
245 experimental quantification of muscle co-contraction is needed to determine the optimal cost function.
246 However, the non-uniform contraction pattern of GM during abduction cannot be predicted using the
247 cost function of minimizing the maximum active contractile stress. It should be noted that co-
248 contraction from antagonistic muscles or antagonistic muscle subdivisions is not considered in these
249 cost functions of optimization. A higher level of contractions in all the subdivisions could be expected
250 in a model considering co-contraction. Although co-contraction would offer similar qualitative
251 predictions regarding the trend of results and the comparison between models, it will be considered
252 in future studies once reliable experimental quantification of muscle co-contraction is available.

253 Obviously, there were a number of limitations in this study. First, the non-uniform contraction
254 that could further exist in each GM subdivision would provide smoother distribution of stresses and
255 strains across the muscle compared to the current model assuming uniform contraction in each
256 subdivision. This aspect was not accounted for in the current model, since such studies are not
257 available yet for comparison. Secondly, it was assumed that motor neurons mainly stimulate active
258 stresses of muscles, so only active stresses were optimized in this study. However, passive stresses
259 and strains would also play a role in the muscle recruitment pattern, which will be a future
260 consideration. Additionally, the material properties of the muscle were adopted from the literature.
261 Variation in material properties among individuals and muscles should be considered for future
262 subject-specific studies. Constitutive models accounting for the effect of fibre stretches and velocity
263 on the level of muscle activation and derived specifically for modelling of muscles (e.g. coupling of
264 active and passive stresses) [37-39] will also be a consideration in future studies. Although the
265 modelling and optimization procedures are based on a validated algorithm [26, 40], the GM model
266 itself was not validated due to the challenge in experimental measurements of the biomechanical
267 behaviour of GM. Future more reliable experimental measurements, perhaps through dynamic
268 imaging [41, 42], would provide alternative data to validate the change in morphology of the muscle
269 model due to contraction. In the current model, the other abductor and rotator muscles of the hip were
270 not accounted for, since the purpose of this study was to evaluate the activity of GM performing
271 various rotation tasks under constant moments. However, the surrounding tissues would have an
272 effect on the biomechanical behaviour of the muscle [26]. Apart from the motions simulated in this
273 study, other motions including flexion, extension and adduction were not investigated, because GM
274 was found to play a minor role in these motions. The muscle model with non-uniform contraction
275 developed in this study will be incorporated into a 3D FE musculoskeletal modelling framework we
276 have developed recently [26] to enable more realistic musculoskeletal modelling and simulations of
277 a wider range of activities such as gait in future work. However, these assumptions do not affect the

278 qualitative predictions of this study, i.e. biomechanical differences found between models during
279 different motions and with different cost functions and dimensions of subdivisions.

280 In conclusion, a 3D FE model of the GM muscle incorporating inverse-dynamics-based
281 optimization was developed to simulate non-uniform contraction of the muscle. Contraction across
282 GM was found to be highly non-uniform during various motions. The whole GM was activated during
283 abduction, whereas only the anterior and posterior subdivisions were primarily involved in internal
284 and external rotation, respectively. The proportion in the dimension of each GM subdivision had an
285 effect on the predictions and therefore may be important to consider for subject-specific modelling
286 of non-uniform muscle contraction. The cost functions of optimization that have been widely used in
287 previous musculoskeletal models of the lower extremity, i.e. minimizing the sum of active contractile
288 stresses squared/cubed, provide similar qualitative predictions of muscle activity. This computational
289 approach has the potential to aid in understanding the mechanisms of muscle function and the
290 pathology of musculoskeletal impairments. Future attempts will also be made to simulate the electro-
291 mechanics behaviour of muscles using the modelling and optimization framework.

292

293 **Acknowledgements**

294 This research was supported by the Newton Fund of the British Council 2017-RLWK9-10075 and
295 the Open Fund from the China State Key Laboratory of Structural Analysis for Industrial Equipment
296 (GZ19108). The author appreciates the efforts of the developers of FEBio.

297 **References**

- 298 1. Brandt, M., et al., *Participatory intervention with objectively measured physical risk factors for*
299 *musculoskeletal disorders in the construction industry: study protocol for a cluster randomized*
300 *controlled trial*. BMC Musculoskeletal Disorders, 2015. **16**(1): p. 302.
- 301 2. Golabchi, F.N., et al., *Assessing aberrant muscle activity patterns via the analysis of surface EMG data*
302 *collected during a functional evaluation*. BMC musculoskeletal disorders, 2019. **20**(1): p. 13-13.
- 303 3. Niemuth, P.E., et al., *Hip muscle weakness and overuse injuries in recreational runners*. Clinical
304 Journal of Sport Medicine, 2005. **15**(1): p. 14-21.
- 305 4. Nakagawa, T.H., et al., *Trunk, pelvis, hip, and knee kinematics, hip strength, and gluteal muscle*
306 *activation during a single-leg squat in males and females with and without patellofemoral pain*
307 *syndrome*. Journal of Orthopaedic & Sports Physical Therapy, 2012. **42**(6): p. 491-501.
- 308 5. Gizzi, A., et al., *Theoretical and Numerical Modeling of Nonlinear Electromechanics with applications*
309 *to Biological Active Media*. Communications in Computational Physics, 2014. **17**(1): p. 93-126.
- 310 6. Klotz, T., et al., *Modelling the electrical activity of skeletal muscle tissue using a multi-domain*
311 *approach*. Biomechanics and Modeling in Mechanobiology, 2020. **19**(1): p. 335-349.
- 312 7. Loppini, A., et al., *Competing Mechanisms of Stress-Assisted Diffusivity and Stretch-Activated*
313 *Currents in Cardiac Electromechanics*. Frontiers in Physiology, 2018. **9**(1714).
- 314 8. Semciw, A.I., et al., *Gluteus medius: An intramuscular EMG investigation of anterior, middle and*
315 *posterior segments during gait*. Journal of Electromyography and Kinesiology, 2013. **23**(4): p. 858-
316 864.
- 317 9. Watanabe, K., M. Kouzaki, and T. Moritani, *Non-uniform surface electromyographic responses to*
318 *change in joint angle within rectus femoris muscle*. Muscle & Nerve, 2014. **50**(5): p. 794-802.
- 319 10. Gottschalk, F., S. Kourosh, and B. Leveau, *The functional anatomy of tensor fasciae latae and gluteus*
320 *medius and minimus*. Journal of anatomy, 1989. **166**: p. 179-189.
- 321 11. O'Sullivan, K., S.M. Smith, and D. Sainsbury, *Electromyographic analysis of the three subdivisions of*
322 *gluteus medius during weight-bearing exercises*. BMC Sports Science, Medicine and Rehabilitation,
323 2010. **2**(1): p. 17.
- 324 12. Soderberg, G.L. and W.F. Dostal, *Electromyographic study of three parts of the gluteus medius muscle*
325 *during functional activities*. Physical therapy, 1978. **58**(6): p. 691-696.
- 326 13. O'Dwyer, C., D. Sainsbury, and K. O'Sullivan, *Gluteus medius muscle activation during isometric*
327 *muscle contractions*. Journal of Sport Rehabilitation, 2011. **20**(2): p. 174-186.
- 328 14. Kolk, S., et al., *Symmetry and spatial distribution of muscle glucose uptake in the lower limbs during*
329 *walking measured using FDG-PET*. PLOS ONE, 2019. **14**(4): p. e0215276.
- 330 15. Goodwin, P.C., et al., *Reliability of leg muscle electromyography in vertical jumping*. European Journal
331 of Applied Physiology and Occupational Physiology, 1999. **79**(4): p. 374-378.
- 332 16. Ghaderi, P. and H.R. Marateb, *Muscle activity map reconstruction from high density surface EMG*
333 *signals with missing channels using image inpainting and surface reconstruction methods*. IEEE
334 Transactions on Biomedical Engineering, 2017. **64**(7): p. 1513-1523.
- 335 17. Young, C.C., et al., *The effect of surface and internal electrodes on the gait of children with cerebral*
336 *palsy, spastic diplegic type*. Journal of Orthopaedic Research, 1989. **7**(5): p. 732-737.
- 337 18. Barbosa, A.C., et al., *Increased activation amplitude levels of gluteus medius in women during*
338 *isometric and dynamic conditions following a 4-week protocol of low-load eccentric exercises*.
339 Physiotherapy Research International, 2016. **21**(4): p. 257-263.
- 340 19. Chen, Z., et al., *Effect of component mal-rotation on knee loading in total knee arthroplasty using*
341 *multi-body dynamics modeling under a simulated walking gait*. Journal of Orthopaedic Research,
342 2015. **33**(9): p. 1287-1296.
- 343 20. Li, J., et al., *Unilateral total hip replacement patients with symptomatic leg length inequality have*
344 *abnormal hip biomechanics during walking*. Clinical Biomechanics 2015. **30**(5): p. 513-519.
- 345 21. Webb, J.D., S.S. Blemker, and S.L. Delp, *3D finite element models of shoulder muscles for computing*
346 *lines of actions and moment arms*. Computer Methods in Biomechanics and Biomedical Engineering,
347 2014. **17**(8): p. 829-837.

22. Zöllner, A.M., et al., *On high heels and short muscles: A multiscale model for sarcomere loss in the gastrocnemius muscle*. Journal of Theoretical Biology, 2015. **365**: p. 301-310.
23. Stelletta, J., R. Dumas, and Y. Lafon, *Modeling of the Thigh: A 3D Deformable Approach Considering Muscle Interactions*, in *Biomechanics of Living Organs*, Y. Payan and J. Ohayon, Editors. 2017, Academic Press: Oxford. p. 497-521.
24. Mo, F., et al., *A lower limb-pelvis finite element model with 3D active muscles*. Annals of Biomedical Engineering, 2018. **46**(1): p. 86-96.
25. Mo, F., et al., *Implementation of controlling strategy in a biomechanical lower limb model with active muscles for coupling multibody dynamics and finite element analysis*. Journal of Biomechanics, 2019. **91**: p. 51-60.
26. Li, J., et al., *Development of a finite element musculoskeletal model with the ability to predict contractions of three-dimensional muscles*. Journal of Biomechanics, 2019. **94**: p. 230-234.
27. Carbone, V., et al., *TLEM 2.0 – A comprehensive musculoskeletal geometry dataset for subject-specific modeling of lower extremity*. Journal of Biomechanics, 2015. **48**(5): p. 734-741.
28. Li, J., et al., *The influence of the representation of collagen fibre organisation on the cartilage contact mechanics of the hip joint*. Journal of Biomechanics, 2016. **49**(9): p. 1679-1685.
29. Weiss, J.A., B.N. Maker, and S. Govindjee, *Finite element implementation of incompressible, transversely isotropic hyperelasticity*. Computer Methods in Applied Mechanics and Engineering, 1996. **135**(1): p. 107-128.
30. Tsutsumi, M., A. Nimura, and K. Akita, *The gluteus medius tendon and its insertion sites: an anatomical study with possible implications for gluteus medius tears*. JBJS, 2019. **101**(2): p. 177-184.
31. Maas, S.A. and J.A. Weiss, *FEBio Theory Manual*. 2007: <http://mrl.sci.utah.edu/software/febio>.
32. Heintz, S. and E.M. Gutierrez-Farewik, *Static optimization of muscle forces during gait in comparison to EMG-to-force processing approach*. Gait and Posture, 2007. **26**(2): p. 279-288.
33. Adouni, M. and A. Shirazi-Adl, *Partitioning of knee joint internal forces in gait is dictated by the knee adduction angle and not by the knee adduction moment*. Journal of Biomechanics, 2014. **47**(7): p. 1696-1703.
34. Rasmussen, J., M. Damsgaard, and M. Voigt, *Muscle recruitment by the min/max criterion — a comparative numerical study*. Journal of Biomechanics, 2001. **34**(3): p. 409-415.
35. Dul, J., et al., *Muscular synergism—I. On criteria for load sharing between synergistic muscles*. Journal of Biomechanics, 1984. **17**(9): p. 663-673.
36. Li, J., et al., *Hip contact forces in asymptomatic total hip replacement patients differ from normal healthy individuals: Implications for preclinical testing*. Clinical Biomechanics 2014(0).
37. Gizzi, A., M. Vasta, and A. Pandolfi, *Modeling collagen recruitment in hyperelastic bio-material models with statistical distribution of the fiber orientation*. International Journal of Engineering Science, 2014. **78**: p. 48-60.
38. Vasta, M., A. Gizzi, and A. Pandolfi, *A spectral decomposition approach for the mechanical statistical characterization of distributed fiber-reinforced tissues*. International Journal of Non-Linear Mechanics, 2018. **106**: p. 258-265.
39. Röhrle, O., J.B. Davidson, and A.J. Pullan, *A physiologically based, multi-scale model of skeletal muscle structure and function*. Frontiers in physiology, 2012. **3**: p. 358-358.
40. Li, J., *Development and validation of a finite-element musculoskeletal model incorporating a deformable contact model of the hip joint during gait*. Journal of the Mechanical Behavior of Biomedical Materials, 2021. **113**: p. 104136.
41. Sinha, S., et al., *Muscle kinematics during isometric contraction: Development of phase contrast and spin tag techniques to study healthy and atrophied muscles*. Journal of Magnetic Resonance Imaging, 2004. **20**(6): p. 1008-1019.
42. Mazzoli, V., et al., *Accelerated 4D phase contrast MRI in skeletal muscle contraction*. Magnetic Resonance in Medicine, 2018. **80**(5): p. 1799-1811.
43. Blemker, S.S., P.M. Pinsky, and S.L. Delp, *A 3D model of muscle reveals the causes of nonuniform strains in the biceps brachii*. Journal of Biomechanics, 2005. **38**(4): p. 657-665.
44. Teran, J., et al., *Creating and simulating skeletal muscle from the visible human data set*. IEEE Transactions on Visualization and Computer Graphics, 2005. **11**(3): p. 317-328.

- 401 45. Röhrle, O. and A.J. Pullan, *Three-dimensional finite element modelling of muscle forces during*
402 *mastication*. Journal of Biomechanics, 2007. **40**(15): p. 3363-3372.
403

Figure Legends

Fig. 1. a – FE model of the GM muscle in the original configuration (posterior-lateral view); b – fibre architecture in the tissue as illustrated by the blue arrows.

Fig. 2. GM models with subdivisions of varying dimensions in the original configuration (posterior-lateral view; anterior, medial and posterior subdivisions in blue, yellow and red, respectively; tendons in orange). a – GM subdivisions of similar dimension; b, c and d – one of the three GM subdivisions approximately 1.5 times larger than the other two portions.

Fig. 3. Contour of tensile stresses in the GM model in the original configuration (0 degrees rotation) and at 20 degrees rotation during abduction, internal rotation and external rotation (posterior-lateral view), showing stress distribution in GM under non-uniform contraction. The cost function was minimizing the sum of active contractile stresses cubed. The three subdivisions were of similar dimensions with boundaries marked by the black lines. Tensile stresses were concentrated around the tissue connecting the femur. Tensile stresses in all the subdivisions were of similar levels during hip abduction. Tensile stresses were markedly higher in the anterior subdivision and posterior subdivision during internal rotation and external rotation, respectively.

Fig. 4. Active contractile stresses in the GM subdivisions during abduction, internal rotation and external rotation. The three subdivisions were of similar dimensions. The cost function was minimizing the sum of active contractile stresses cubed. Active contractile stresses increased with increasing rotation angle during all the motions. All the subdivisions were activated during abduction, while only the anterior subdivision was activated during internal rotation. During external rotation, both the medial and posterior subdivisions were activated at 0 degrees and 5 degrees, while the medial subdivision became inactivated when the rotation exceeded 10 degrees. During abduction, active contractile stresses of the anterior and medial subdivisions were similar and higher than the posterior portion.

Fig. 5. Comparison of active contractile stresses during abduction calculated through different cost functions. The three subdivisions were of similar dimensions. The results were compared to the model with uniform contraction across the whole muscle (system not redundant and thus optimization not needed). Results of the models with uniform contraction and with the cost function of minimizing the maximum active contractile stress (min/max) were identical (the two curves overlap). There was a difference in the active contractile stresses under different cost functions, particularly in the posterior subdivision. A higher level of non-uniform contraction was observed under the cost function of minimizing the sum of active contractile stresses squared than minimizing the sum of active contractile stresses cubed. Under the two cost functions of minimizing the sum of active contractile stresses squared/cubed, active contractile stress increased with increasing abduction angle and was higher in the anterior and medial subdivisions than the posterior portion.

Fig. 6. Comparison of active contractile stresses during abduction in the GM with subdivisions of different dimensions. The cost function was minimizing the sum of active contractile stresses cubed. There was a marked difference in active contractile stresses among the models with subdivisions of different dimensions. The active contractile stress was higher in a larger subdivision compared with the same subdivision of the model with subdivisions of similar dimensions.

Fig 1

Fig 2

Fig 3

Fig 4

Fig 5

Fig 6

List of Tables

Table 1. Constitutive model parameters. The values are based on previous studies [24, 43-45]. The maximum value of active contractile stress was 0.5 MPa.

Muscle constants		Tendon constants	
C_1	0.01 MPa	C_1	0.1 MPa
C_2	0.01 MPa	C_2	0.1 MPa
C_3	0.015 MPa	C_3	2.7 MPa
C_4	15	C_4	46.4
C_5	6 MPa	C_5	500 MPa
K	10 MPa	K	100 MPa
λ_m	1.4	λ_m	1.03