ESTIMATION OF HIP-MUSCLE GEOMETRY USING AUTOMATED, NON-RIGID ATLAS-BASED REGISTRATION OF MR IMAGES

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INTRODUCTION

The analysis of parameters related to the muscle-tendinous structures (such as muscle tendon length, moment arm) during gait is highly relevant in patients presenting motor coordination dysfunction as well as skeletal deformity. Such analysis requires the use of musculoskeletal (MS) models. In general, these models are created by rescaling and deforming generic models based on data of healthy adult men. However fast, this procedure leads to models suffering from low accuracy and little patient-specific detail. Especially when analyzing gait in a paediatric and/or pathologic population, accommodation of the models for inter-individual variability in musculoskeletal geometry is needed. Modelling based on Magnetic Resonance Imaging (MRI) has great potential to generate individualized MS models. However, at present, the level of user interaction required for parameter extraction based on manual segmentation limits the applicability of this approach in a broader context.

This work presents the potential of a method to automatically identify muscle geometry (attachments and muscle path) by atlas-based non-rigid image registration in patients with increased femoral anteversion (FA).

METHODS AND PATIENTS

A pediatric atlas (age 9y, FA 31˚) was build by manual delineation of muscle geometry in a full-leg MRI volume, using a custom build software tool [ref]. Based on this atlas, muscle paths were automatically identified in MR images from 3 cerebral palsy (CP) subjects (average age 10.3y, average FA 38˚) using non-rigid image registration: in a first step the geometrical relation between both image volumes was determined by matching the image volumes using intensity-based, non-rigid image registration. The non-rigid deformation of the atlas image was modelled by a B-spline deformation mesh. The cost function uses mutual information as a similarity measure while a volume penalty term discourages improbable or impossible deformations. In a second step, the patient's muscle paths were retrieved by applying the resulting 3D deformation field to the original musculoskeletal geometry from the atlas. For each subject, a reference model was build using manual delineation. Hip joint centres were calculated by fitting a sphere to the segmented femoral head, using the iterative closest point algorithm.

Absolute differences in musculotendon (MTL) and moment arm lengths (MAL) for 21 major hip muscles are calculated between (1) the automatically and (2) the manually defined MS model using SIMM (Musculographics, Inc.). This was done for the scanned supine position. For MTL, differences are normalized to the MTL of the manually defined MS model. For MAL, absolute differences are reported for each plane of movement (adduction, rotation and flexion).

RESULTS

Table 1 presents the differences in MTL and MAL for the 21 muscles in the model. The average differences in normalized MTL remain below 10% in each of the subjects, with a maximum over 20% for the M. adductor longus and brevis. Larger MTL-errors are present in muscles with two or more lines of action (e.g. adductor magnus and glutei). The average error in these muscles measured 10.8, 11.1 and 10.8 compared to 6.7, 5.5 and 7.0 for each subject respectively. Average errors in MAL range between 3.51 mm in hip rotation, 4.62 mm in hip adduction and 6.68 mm in flexion, with maximum errors just over 20 mm reported in M. Gracilis for flexion and M. Adductor brevis for adduction.

DISCUSSION

Comparison of muscle tendon lengths and muscle moment arm lengths between the automatically and manually defined MS models shows that non-rigid, atlas-based registration of the MR images results in a valid estimate of the muscle paths, even in the presence of a large variation in femoral geometry. The proposed method therefore yields a good initial guess for user evaluation of the automatically proposed muscle line of action combined with the MR images. Consequently, the amount of user interaction as well as the total modelling time required for the creation of subject-specific musculoskeletal models can be reduced substantially. Careful selection of the appropriate atlas image will be essential to further improve the performance of the presented methodology and test the generalization of its performance in other pathological conditions.

SUMMARY/CONCLUSIONS

Automatic identification of muscle geometry using atlas-based, non-rigid registration of MR images enhances the feasibility to build detailed, subject-specific musculoskeletal models for biomechanical analysis of gait deviations in subjects suffering aberrant musculoskeletal geometry.

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Table 1: calculated absolute differences between manual and automatic delineation for all muscles from subject A (age 8, FA 42°,, ♂), B (age 11, FA 27°, ♂) and C (age 12, FA 46°, ♀).

	Hip Adduction MAL (mm)			**Hip Rotation MAL (mm)**			**Hip Flexion MAL (mm)**			**Normalized MTL**		
Subject =>	A	B	C	A	B	C	A	B	C	A	B	C
Glut max (n=3)	6.9	5.4	4.6	7.4	4.1	3.4	2.4	8.4	3.2	10.1	11.7	12.5
Glut med (n=3)	1.5	1.8	3.6	3.5	6.3	3.4	6.4	3.2	5.1	12.1	12.3	22.5
Glut min (n=3)	2.4	2.7	3.8	7.1	7.9	3.4	1.5	0.7	8.4	8.3	13.5	4.9
Add mag (n=3)	14.3	0.6	3.2	1.6	2.7	4.1	14.4	14.5	10.7	12.5	6.8	3.0
Tensor FL	3.5	2.5	5.8	5.5	4.6	8.3	1.7	8.5	0.7	3.4	6.8	2.6
Add long	11.7	0.2	6.4	3.6	3.6	9.2	3.8	11.6	3.9	7.3	26.7	24.8
Add brev	21.1	7.9	5.6	7.4	4.0	0.4	8.4	17.5	2.6	24.8	11.0	10.1
Gracilis	11.3	2.7	13.9	1.6	1.3	1.2	18.3	27.1	8.6	3.2	0.3	8.2
Semimem	2.4	0.4	1.3	0.4	0.5	0.7	4.5	10.5	0.4	2.1	0.2	0.2
Semiten	1.8	3.1	3.3	1.5	9.4	2.6	5.0	8.4	3.8	4.8	2.0	0.2
Bi Fem LH	0.8	3.8	2.9	0.5	0.2	0.8	1.5	8.4	6.6	3.8	0.2	6.7
Sartorius	2.8	3.2	2.4	3.2	3.3	4.8	4.9	7.1	3.0	4.2	2.3	4.7
Rect fem	3.1	3.5	2.2	1.5	0.2	1.5	3.4	1.2	0.5	6.6	0.2	5.7
Average	**4.62 ± 4.4**			**3.51 ± 2.69**			**6.68 ± 5.73**			**7.78 ± 7.07**		