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Effect of sub-optimal neuromotor control on the hip joint load during level walking

Saulo Martelli [a,*], Fulvia Taddei [a], Angelo Cappello [b], Serge van Sint Jan [c], Alberto Leardini [d], Marco Viceconti [a]

[a] Laboratorio di Tecnologia Medica, Istituto Ortopedico Rizzoli, Bologna, Italy
[b] Dipartimento di Elettronica, Informatica e Sistemistica, Università di Bologna, Italy
[c] Department of Anatomy, Université Libre de Bruxelles, Belgium
[d] Movement Analysis Laboratory, Istituto Ortopedico Rizzoli, Bologna, Italy

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A B S T R A C T

Skeletal forces are fundamental information in predicting the risk of bone fracture. The neuromotor control system can drive muscle forces with various task- and health-dependent strategies but current modelling techniques provide a single optimal solution of the muscle load sharing problem. The aim of the present work was to study the variability of the hip load magnitude due to sub-optimal neuromotor control strategies using a subject-specific musculoskeletal model. The model was generated from computed tomography (CT) and dissection data from a single cadaver. Gait kinematics, ground forces and electromyographic (EMG) signals were recorded on a body-matched volunteer. Model results were validated by comparing the traditional optimisation solution with the published hip load measurements and the recorded EMG signals. The solution space of the instantaneous equilibrium problem during the first hip load peak resulted in 10^5 dynamically equivalent configurations of the neuromotor control. The hip load magnitude was computed and expressed in multiples of the body weight (BW). Sensitivity of the hip load boundaries to the uncertainty on the muscle tetanic stress (TMS) was also addressed. The optimal neuromotor control induced a hip load magnitude of 3.3 BW. Sub-optimal neuromotor controls induced a hip load magnitude up to 8.93 BW. Reducing TMS from the maximum to the minimum the lower boundary of the hip load magnitude varied moderately whereas the upper boundary varied considerably from 4.26 to 8.93 BW. Further studies are necessary to assess how far the neuromotor control can degrade from the optimal activation pattern and to understand which sub-optimal controls are clinically plausible. However we can consider the possibility that sub-optimal activations of the muscular system play a role in spontaneous fractures not associated with falls.

1. Introduction

Load cells and force platforms make possible accurate measurement of the external forces that act on our body during activities of daily living (Sutherland, 2005). However, the determination of the internal forces that the same physical activity induces on our skeleton through the joints, the ligaments, and the muscle insertions remain difficult to quantify. But this information is of vital importance in a number of research and clinical contexts. For example, the risk of fracture that a given subject faces while performing a given motor task depends not only on the specific bone strength, but also on the internal forces.

The problem is affected by a dramatic indeterminacy. Even if we model the skeleton as a mechanism made of idealised joints, represent each major muscle bundle with a single actuator, and impose all the physiological limits to the force expressed by each actuator, the resulting mathematical problem has more unknowns than equations. The best solution, when the kinematics of each segment has been measured experimentally, is to postulate that the neuromotor control activates the muscle fibres ensuring the instantaneous equilibrium while minimising a cost function (Collins, 1995; Menegaldo et al., 2006; Praagman et al., 2006).

The assumption that in healthy subjects the neuromotor control works in fairly optimal conditions seems reasonable. Indeed, when applied to volunteers this approach predicts muscle activation patterns in good agreement with electromyography (EMG) recordings (Anderson and Pandy, 2001; Erdemir et al., 2007; Heller et al., 2001). Also, the intensity of the hip load predicted is comparable to that recorded with telemetric instrumented prostheses (Heller et al., 2001; Stansfield et al., 2003).

* Correspondence to: Laboratorio di Tecnologia Medica, Istituto Ortopedico Rizzoli, Via di Barbiano, 1/10, 40136 Bologna, Italy. Tel.: +39 051 6366554; fax: +39 051 6366863.
E-mail address: martelli@tecno.ior.it (S. Martelli).

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This approach presumes that the neuromotor control chooses, among the infinite solutions available, the muscle activation pattern that optimises a certain cost function, always the same one. But this assumption seems unrealistic for the following cases:

- A large variability of the internal forces can be observed for a single subject through several repetitions of the same motion task (Bergmann et al., 2001). The "UnControlled Manifold" theory (Scholz and Schöner, 1999) suggests that the motor control strategy focusses on the goal of the task, and that every trajectory within the manifold of the task-equivalent configuration of the muscle actuators is virtually possible.
- While we move, our goal is dependent on a number of factors. Specific activation patterns were found in case of patello-femoral pain (Besier et al., 2009), in unstable conditions (Bergmann et al., 2004), in sudden motion tasks (Yeadon et al., 2010) and different muscles controls were found during the execution of precise and power activities (Anson et al., 2002).
- The way we move is also affected by our emotions. Depression has been found a co-factor for the risk of falling in the elders (Skelton and Todd, 2007; Talkowski et al., 2008), whereas somatisation, anxiety and depression were found intrinsic co-factors in non-specific musculoskeletal spinal disorders (Andersson, 1999).
- Even if the optimal control assumption is acceptable for normal subjects, does it remain acceptable when used to model specific patients that are known to have neuromotor deficiencies? EMG studies seem to suggest the answer is "no" (Liikavainio et al., 2009; Mahaudens et al., 2009).

More specifically, in predicting the risk of fracture of an elder who has a history of falls, is it still reasonable to postulate that the muscle activation follows some optimal patterns? This question is very difficult to answer in such general terms; but as a first step a simpler question might be addressed: is the intensity of the loads acting on the skeleton during a given motor task severely affected by neuromotor control strategy?

The aim of the present study is to estimate, using a subject-specific musculoskeletal model, how the intensity of the hip joint reaction acting is affected when the instantaneous equilibrium is achieved in full respect of the physiological constraints, but with sub-optimal activation patterns. Sensitivity of potential sub-optimal control strategies to the muscle mechanics parameters was also assessed.

2. Materials and methods

2.1. Data collection

A musculoskeletal model of the lower body was developed from a detailed multiscale data collection (Testi et al., 2010). In this project the cadaver of a 81 years old woman, 167 cm height and 63 kg mass with no history of musculoskeletal diseases was examined with a whole body high-resolution computed tomography (CT) protocol. The muscle attachment on bones, the muscle length and their superficial lines of action were digitised during dissection. The volume of each muscle was measured by water immersion. The co-ordinates of the muscle attachments were registered in space through a dedicated software (LhpBuilder, SCS, Italy) into a unique multimodal data collection.

A body-matched healthy volunteer (female, 25 years old, 57 kg, 165 cm) was subjected to a detailed gait analysis protocol during level walking (Leardini et al., 2007; Manca et al., 2010), which provided 3D motion (Vicon Motion Capture, Oxford UK) of the lower limb segments (sampling rate 100 Hz) and the ground reaction forces at both feet (sampling rate 2000 Hz). The muscle activity was recorded for the most relevant muscles of the lower limb: gluteus maximus, gluteus medius, rectus femoris, vastus lateralis, vastus medialis, biceps femoris long head, semitendinosus, gastrocnemius medialis, gastrocnemius lateralis, soleus and tibialis anterior. Surface EMG signals were acquired (TelEMG®, BTS, Italy) at the sampling rate of 2000 Hz. Digital band filtering was used to extract the qualitative muscle control pattern (Glitsch and Baumann, 1997).

2.2. The musculoskeletal model

The skeletal anatomy was extracted from the CT dataset using dedicated semi-automatic software (Amira®, Mercury Computer System, Inc., USA). The biomechanical model of the musculoskeletal system of the lower-limb was defined as a 7-segment, 10-degree-of-freedom (DOF) articulated system actuated by 82 muscle–tendon units. Each leg was articulated by three ideal joints: a ball and socket at the hip (3 DOF) and a hinge (1 DOF) at both the knee and the ankle (Jonkers et al., 2008). The identification of joint parameters was based on relevant skeletal landmarks identified on the skeletal surface (Taddei et al., 2007). All anatomical landmarks suggested by ISB standards were identified. The local coordinate system was computed for each segment (Wu et al., 2002). The hip centre was defined as the centre of the sphere that best fit the femoral head surface. The hip joint rotations were defined according to ISB standards (Wu et al., 2002). The knee rotation axis was assumed as that connecting the medial and the lateral epicondyles, which is normally considered a good approximation of this axis in surgical procedures (Tanavalee et al., 2001). The ankle flexion axis was assumed as that connecting the medial and the lateral malleoli. An expert anatomist manually registered on the subject-specific skeletal anatomy the muscular model of the lower extremity (Delp et al., 1990) using as reference the muscle line-of-actions digitised during dissection (Fig. 1b). The peak force each muscle can exert was estimated from the physiological cross section area (PCSA), assuming the muscle tetanic stress (TMS) equal to 1 MPa (Glitsch and Baumann, 1997). The muscle mechanical properties were defined accordingly (Daggfeldt and Thorstensson, 2003). Inertial parameters of each segment were derived from CT data (Fig. 1a) assuming homogeneous density properties for both the hard (1.42 g/cm^3) and the soft (1.03 g/cm^3) tissues (Dumas et al., 2005). Preliminary simulations were run to solve the muscle load sharing problem using a traditional static optimisation approach (Anderson and Pandy, 2001; Collins, 1995; Crowninshield and Brand, 1981) (Fig. 1c). The hip load magnitude was calculated and expressed in multiples of the ground reaction peak (GRp) to account for the inter-subject weight and walking dynamics variability.

2.3. Musculoskeletal model validation

The predicted hip load under the hypothesis of optimal neuromotor control was compared to those measured in various subjects using telemetric implants (Bergmann et al., 2001). The predicted force patterns exerted by the principal muscles were compared with the corresponding electrical activity (Glitsch and Baumann, 1997) recorded on the body-matched volunteer during the same motion task. The model configuration that produced the first load peak was identified as the subject for the subsequent analysis of sub-optimal neuromotor control conditions.

2.4. Variability of the hip load in sub-optimal motor-control strategies

The instantaneous equilibrium at the joints is achieved by every set of muscle forces that satisfies the following equation:

$$\overline{\overline{B}}_{m,n} \cdot \overline{F}_n = \overline{M}_m \tag{1}$$

where the matrix $\overline{\overline{B}}_{m,n}$ contains the muscle lever arm of each of the n muscle action lines acting on each of the m DOFs, $\overline{F}_n$ is a n dimension vector containing the design variables, i.e. the muscle forces, and $\overline{M}_m$ is the m dimension vector that contains the joint net moments necessary to follow the target kinematics. As muscle force we considered only the active component, as the passive forces in the frame of motion here considered are negligible, due to the moderate stretch of the muscle fibre in that position. The active force component depends on a number of factors including the current muscle length, velocity and the activation conditions at the previous frame of motion. These dependencies of the muscle forces are properly described by complex relationships that require a large number of parameters to be identified (e.g. see Thelen et al., 2003) and are often affected by considerable uncertainty. Thus, to include all possible conditions that the neuromotor control system can potentially impose to the musculoskeletal system we constrained the spectrum of possible muscle forces between zero (i.e. completely inactive muscle) and the tetanic muscle forces at their optimal length in isometric conditions ($\overline{F}_{\max}$):

$$\overline{F}_n \in [0,\overline{F}_{\max}] \tag{2}$$

This solution space of the non-homogeneous linear system of equations ensuring the instantaneous equilibrium at the joints (Eq. (1)) is given by the null space of the matrix $\overline{\overline{B}}_{m,n}$ plus a particular solution of Eq. (1) (e.g. the optimal neuromotor control solution used in this study). This space is an m–n dimensions hyper-plane in the n-dimensional space of the unknowns (i.e. the muscle forces). In this hyper-plane, the domain of the physiologically plausible muscle forces consist in a