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Technical Note

Identification of passive elastic joint moments in the lower extremities

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Abstract

Musculotendon actuators produce active and passive moments at the joints they span. Due to the existence of bi-articular muscles, the passive elastic joint moments are influenced by the angular positions of adjacent joints. To obtain quantitative information about this passive elastic coupling between lower limb joints, we examined the passive elastic joint properties of the hip, knee, and ankle joint of ten healthy subjects. Passive elastic joint moments were found to considerably depend on the adjacent joint angles. We present a simple mathematical model that describes these properties on the basis of a double-exponential expression. The model can be implemented in biomechanical models of the lower extremities, which are generally used for the simulation of multi-joint movements such as standing-up, walking, running, or jumping. © 1999 Elsevier Science Ltd. All rights reserved.

Keywords: Passive elasticity; Joint properties; Lower extremity; Bi-articular muscles; Joint coupling; Mathematical model

1. Introduction

Several groups have investigated passive elastic joint properties of single joints (Yoon and Mansour, 1982; Audu and Davy, 1985; Esteki and Mansour, 1996; Hatze, 1997). Models that account for these properties have been implemented in biomechanical simulations for the study of gait (Mansour and Audu, 1986; Davy and Audu, 1987), vertical jump (Selbie and Caldwell, 1996), or artificial motor control (Veltink et al., 1992; Riener et al., 1996). However, due to the presence of bi-articular muscles, the passive elastic moment at a joint is influenced by the angular position of the adjacent joint. There is no general mathematical approach known that describes this effect, and little effort has been made to identify this joint-coupling experimentally. Two exceptions are the studies of Yoon and Mansour (1982) and Vrahas et al. (1990), who measured the passive elastic moment at the hip joint and its dependence on the knee joint position. Similarly, Mansour and Audu (1986) recorded the elastic knee joint moment at different adjacent joint positions. Unfortunately, their joint ranges were smaller than the ranges that can be physiologically achieved. To the best of our knowledge, no study has yet investigated the ankle joint moment as a function of knee position.

The goal of this study was to measure the passive elastic joint moment over a wide range of positions of hip, knee, and ankle joint, while also taking into account the influence of the adjacent joint angles. On the basis of these experimental results, a simple mathematical approach is proposed which can be used to estimate the passive elastic joint moments as a function of lower limb angles.

2. Measurements

Moment and angle were measured in ten healthy male subjects ranging in age from 23 to 29 yr (mean weight: 79.8 ± 9.5 kg, mean height: 1.84 ± 0.06 m). All subjects gave their informed consent. Ankle, knee, and hip joint angles were monitored with a special electrogoniometer that is insensitive to irrelevant degrees of freedom such as knee translation (Murray, 1992). A strain-gauge based mono-axial load cell was attached to the limb located distally to the joint being investigated. Angle and force histories were recorded, while an experimenter slowly moved the distal limb in the sagittal plane by pulling or pushing a handle attached to the load cell. During

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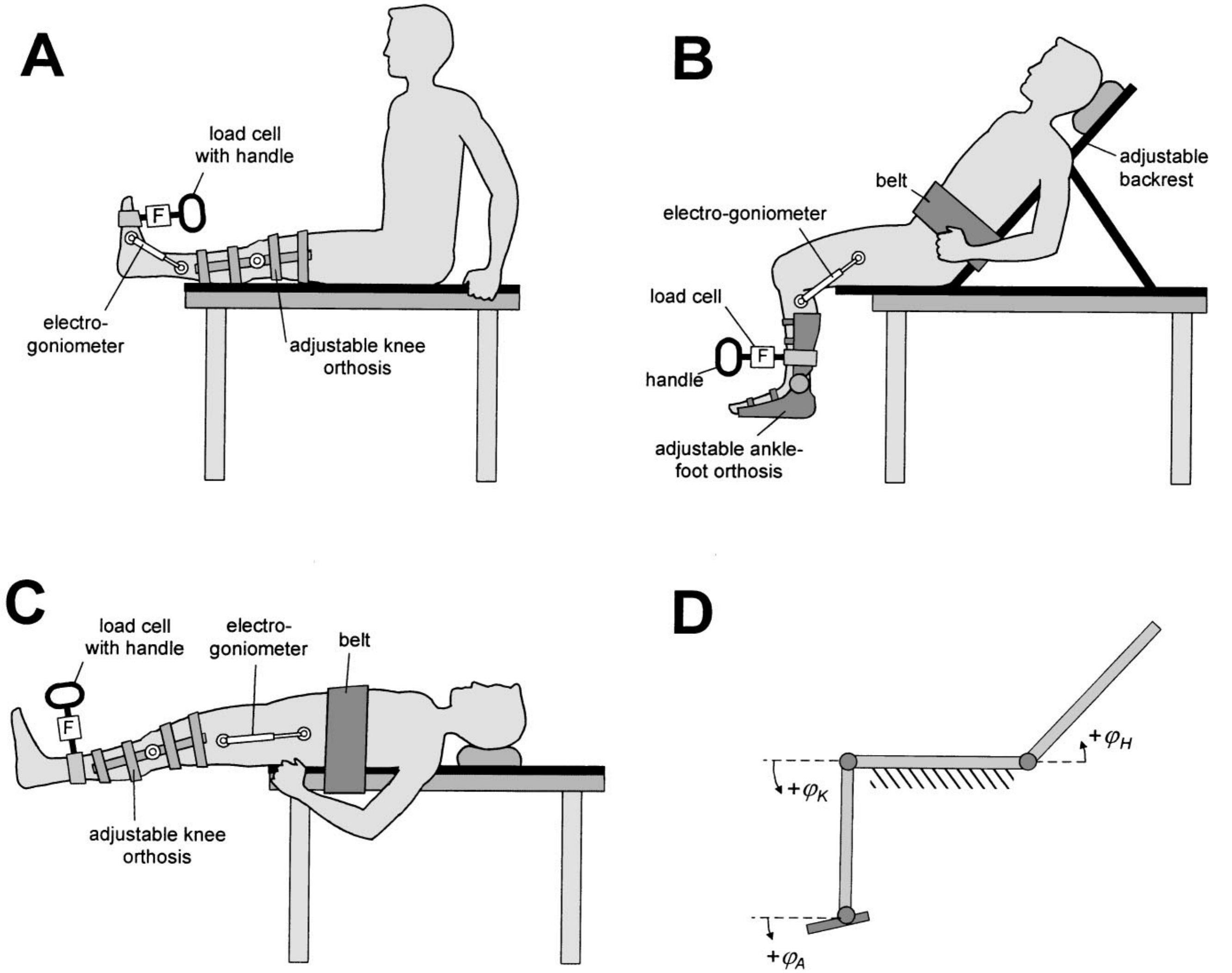

Fig. 1. Experimental setup for measuring the passive joint moments versus joint angles in the ankle (A), knee (B), and hip joint (C). The adjacent joint angles were fixed with an adjustable ankle-foot orthosis, a knee orthosis, or a backrest. The goniometer for measuring joint angles is comprised of two potentiometers connected by a telescopic arm (Murray, 1992). Geometric definitions are illustrated in (D).

recording the adjacent joint positions were held constant (Fig. 1). Data were recorded at (i) the ankle joint with the knee fixed at different knee angles, (ii) the knee joint at different ankle angles, (iii) the knee joint at different hip angles, and (iv) the hip joint at different knee angles.

This experimental setup allowed us to measure over a large range of joint motion. Before data acquisition the limb was cycled in flexion and extension motion, until the subject was accustomed to the motion. Then recording was started, and data for three complete flexion–extension movements were obtained at 10 Hz and low-pass filtered at 1 Hz.

Total joint moment was calculated from the force F measured at the constant distance l to the estimated joint center of rotation. This moment consists of a passive elastic and passive dissipative component (Esteki and Mansour, 1996), a dynamic moment due to inertial effects, and a gravitational moment. The limb was moved very slowly (quasi-statically) so that any dynamic effects are negligible. The gravitational moment can be calculated from the mass m and the center of gravity l_{COG} of limbs and orthosis distal to the joint investigated (Stein et al., 1996). The passive (elastic and dissipative) joint moment can then be approximated by

$$M_{\mathrm{pas}} \approx Fl - mgl_{\mathrm{COG}} \sin(\varphi - \varphi_{\mathrm{r}}), \tag{1}$$

where g is the acceleration of gravity, φ is the joint angle, and φ_{r} is the resting angle at which the leg hangs vertically. The anthropometrical data of the limbs were estimated on the basis of regression equations from Zatsiorski and Seluyanov (1983).

Moment-angle phase plane curves (passive elastic joint moment depicted versus joint angle) are used for further processing and assessment of the recorded data. Upper and lower moment-angle cycles of all ten subjects were averaged to get a representative curve of a ‘generic

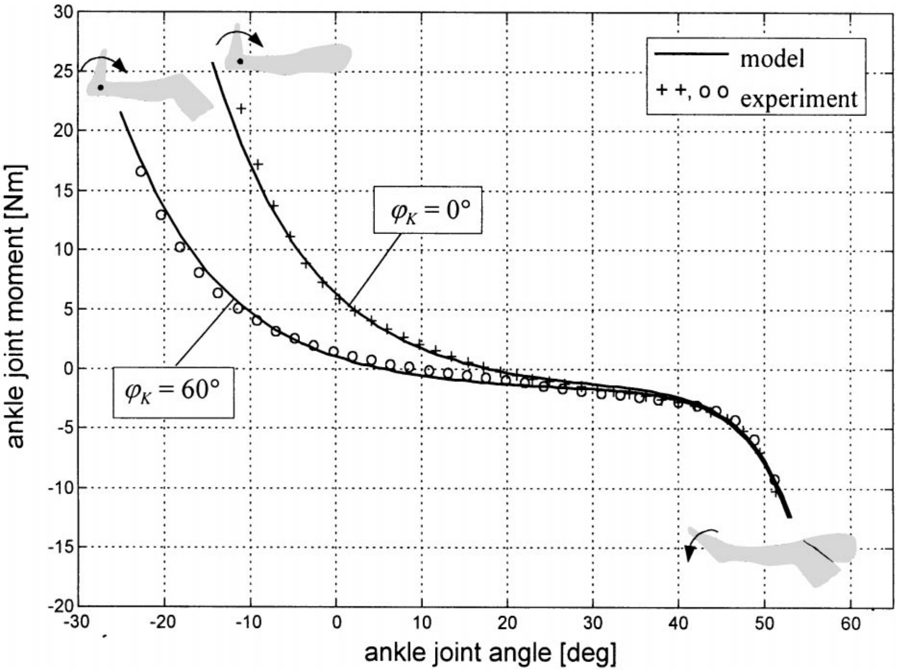

Fig. 2. Measured and predicted ankle joint moment vs. ankle angle at different knee angles. Averaged experimental results at $\varphi_K = 0°$ (+ + +) and $\varphi_K = 60°$ (ooo) as well as the fitted curve from the model (continuous line) are shown.

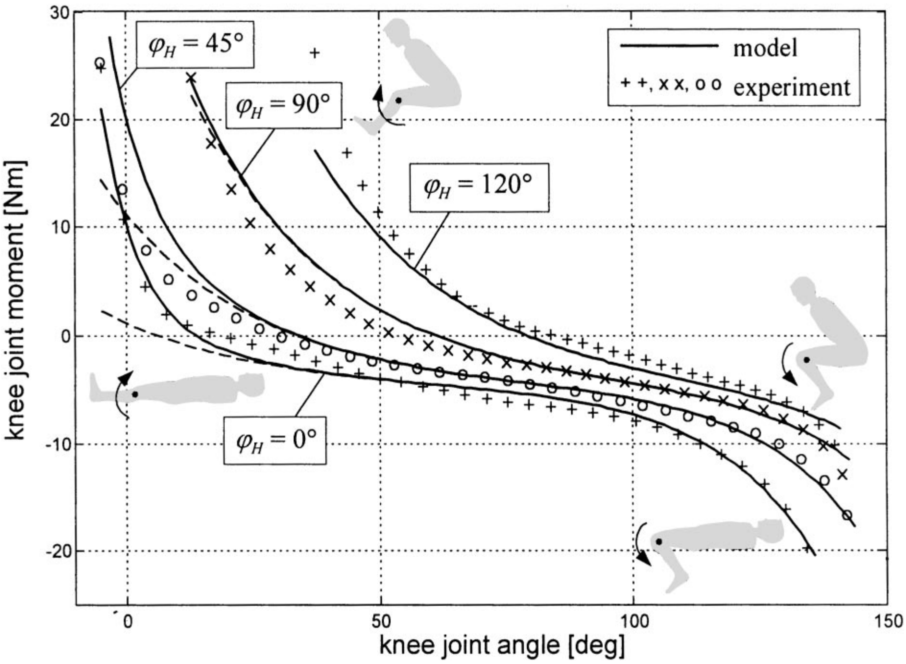

Fig. 3. Measured and predicted knee joint moment vs. knee angle at different hip angles. Averaged experimental results at $\varphi_H = 0°$ (+ + +), $\varphi_H = 45°$ (ooo), $\varphi_H = 90°$ (xxx), and $\varphi_H = 120°$ (+ + +) as well as the fitted curve from the model (continuous line) are shown. The ankle angle was $\varphi_A = 0°$. The dashed line indicates the passive elastic joint moment as modeled without the additional term M_K^* (Eqs. 4 and 6).

subject'. Note that the range of motion of a joint varied significantly among the subjects. Therefore, moment-angle curves were first normalized in angle direction so that all curves shared the identical angle range. Then, moment values were averaged at the same normalized angle values. Finally, the averaged curve was scaled back to the average joint range.

While dissipative effects cause a hysteresis in the moment-angle curves, averaging equal numbers of upper and lower cycles partly compensates for this effect. Thus, the averaged passive moment may be a good approximation for the passive elastic joint moment M_{ela}:

$$M_{ela} \approx M_{pas,\ averaged}. \tag{2}$$

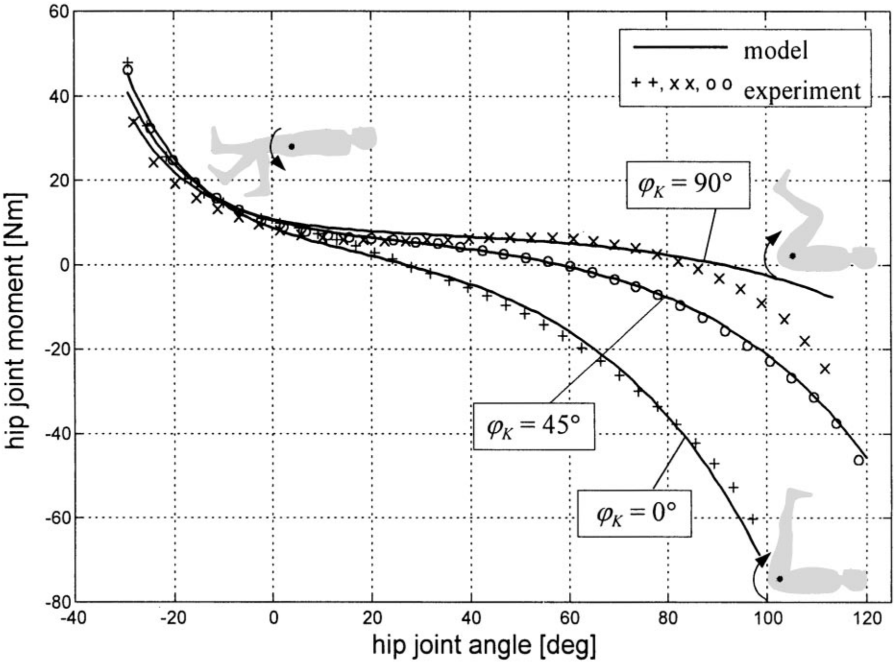

Fig. 4. Measured and predicted hip joint moment vs. hip angle at different knee positions. Averaged experimental results at $\varphi_K = 0°$ (+ + +), $\varphi_K = 45°$ (ooo), and $\varphi_K = 90°$ (xxx) as well as the fitted curve from the model (continuous line) are shown.

The approximation is also justifiable, because joints were rotated slowly to keep the influence of velocity-dependent dissipative effects low (Wright and Johns, 1961; Hatze, 1997). The slight hysteresis effect observed in lower extremity recordings (see Riener and Edrich, 1997; Stein et al., 1996) further indicates that also velocity-independent dissipative effects are considerably low during slow joint movements.

3. Experimental results

Intra-subject variability was tested in two subjects. Recordings on subsequent days showed good repeatability of the measurements. For all joints investigated, the coefficients of variation were approximately equally distributed over the entire joint ranges (mean value for subject A: 5.5%, subject B: 9.1%).

Inter-subject variability in the measured data (e.g. moment and angle range) was considerable. However, the moment-angle curves of the ten subjects all showed the same qualitative behavior. There was a strong correlation between passive elastic joint moments and the positions of the adjacent joints in the lower extremities.

The passive elastic moment of the ankle joint (Fig. 2) depends on the position of the knee joint over a wide range. This can be attributed to the biasing passive elastic moment induced at the ankle by the stretching of the bi-articular gastrocnemius muscle. The dependency disappears for plantarflexion angles $\varphi_A > 20°$, since there is no bi-articular dorsiflexor muscle.

The influence of the hip angle on the elastic knee joint moment is also considerable (Fig. 3). This may be attributed to the moment induced by bi-articular hamstring muscles. The interaction between knee and hip can also be observed in the passive elastic moment data of the hip joint obtained at different knee angles (Fig. 4). In contrast to the passive elastic knee-hip coupling, the interaction between the passive elastic joint moment of the knee and ankle angle is only minor. This can be explained by the fewer bi-articular muscles that span knee and ankle as well as their smaller moment arms (Visser et al., 1990; Spoor and Leeuwen, 1992).

4. Model

On the basis of the so-called double-exponential function (Yoon and Mansour, 1982; Audu and Davy, 1985), a simple model was derived to estimate the passive elastic joint moments of a 'generic subject' as a function of lower limb angles:

$$M_A = \exp(2.1016 - 0.0843\varphi_A - 0.0176\varphi_K) - \exp(-7.9763 + 0.1949\varphi_A + 0.0008\varphi_K) - 1.792. \quad (3)$$

$$M_K = \exp(1.800 - 0.0460\varphi_A - 0.0352\varphi_K + 0.0217\varphi_H) - \exp(-3.971 - 0.0004\varphi_A + 0.0495\varphi_K - 0.0128\varphi_H) - 4.820 + M_K^*. \quad (4)$$

$$M_H = \exp(1.4655 - 0.0034\varphi_K - 0.0750\varphi_H) - \exp(1.3403 - 0.0226\varphi_K + 0.0305\varphi_H) + 8.072, \quad (5)$$

where φ_A, φ_K, and φ_H are ankle, knee, and hip joint angles, respectively, in degrees. Model parameters were determined by fitting the simulated joint moment curves to the averaged measured curves with a least square search procedure provided by MATLAB®.

The model approximates experimental data quite well (Figs. 2–4). Mean deviations between measured and predicted joint moments are rather low (< 2.9 Nm). The best fit is observed for the ankle joint (mean deviation: 0.4 Nm, Fig. 2). For the estimation of the knee joint moment, the largest error occurs at $\varphi_K = 0°$, when the hip is flexed at 45° (maximum deviation 7.5 Nm, Fig. 3). A high deviation is also observed at the hip joint, when the knee is flexed at 90° (> 10 Nm, Fig. 4). There is a general tendency for model error to increase at the extremes of joint ranges.

An additional term M_K^* was introduced to better account for the steep increase of elastic joint moment which develops when the knee is fully extended (Fig. 3):

$$M_K^* = \exp(2.220 - 0.150\varphi_K). \tag{6}$$

Further investigation is necessary to determine whether the sudden increase of the knee joint moment is due to a contribution of the knee joint capsule.

5. Discussion

In this study we demonstrated that there is a strong correlation between the passive elastic joint moments and the joint positions of the adjacent joints in the lower extremities. A truly passive measurement is performed when no reflexes occur and the subjects do not voluntarily contract the muscles. Similar studies have shown that this assumption is true for slow joint speeds (Yoon and Mansour, 1982; Vrahas et al., 1990; Wright and Johns, 1961; Burke et al., 1970; Burke et al., 1971).

Nevertheless, there are several potential sources of errors. Joint angle and moment measurement can be distorted by skin movements, erroneous estimation of limb axes and joint centers, and non-tangential force application by the experimenter. However, the small intra-subject variability indicates that these problems remain within acceptable limits. Furthermore, it can be shown that an angular deviation from the tangential direction does not have any substantial effect on the measured joint moment. For example, a deviation of 10° leads to an overestimation of only 1.54%. Also errors in the joint center estimation do not significantly affect the result after gravity compensation (Eq. (1)). At the ankle gravity almost nothing contributes to the measured moment. At knee and hip the lever arm is so large that the error amounts to only a small percentage of the entire lever arm.

Our simple mathematical model captures passive elastic coupling caused by bi-articular muscles in the lower extremities. It accounts for passive muscle properties that are described as 'total' passive joint moments and are elicited separately from active muscle properties. Other approaches use a musculo-tendon model, in which the passive and the active forces are generated by single muscles (Zajac, 1989; Pandy et al., 1990). However, such models have too many parameters that cannot be identified non-invasively due to the muscle-joint redundancy of the musculoskeletal system. It is easier to consider the passive elastic forces as contributions to the total joint moment and, therefore, measure only the total joint moment. We share the statement made by Zajac and Winters (1990): 'There is thus little reason to distribute passive properties unless individual passive tissue loading, or joint contact loading, is of interest'.

The model presented can be implemented in more complex biomechanical models to simulate movements, e.g. standing-up (Riener and Edrich, 1997), walking, or running. Ignoring passive elastic coupling between adjacent joints can lead to a considerable over- or underestimation of active muscle force and total joint moment, yielding erroneous results in multi-joint movement simulation studies.

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