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Design and Test of a Biomechanical Model for the Estimation of Knee Joint Angle During Indoor Rowing: Implications for FES-Rowing Protocols in Paraplegia

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***Abstract*— Functional electrical stimulation of lower limb muscles during rowing provides a means for the cardiovascular conditioning in paraplegia. The possibility of shaping stimulation profiles according to changes in knee angle, so far conceived as changes in seat position, may help circumventing open issues associated with muscle fatigue and movement coordination. Here we present a subject-specific biomechanical model for the estimation of knee joint angle during indoor rowing. Anthropometric measurements and foot and seat position are inputs to the model. We tested our model on two samples of elite rowers; 15 able-bodied and 11 participants in the Rio 2016 Paralympic games. Paralympic rowers presented minor physical disabilities (LTA-PD classification), enabling them to perform the full rowing cycle (with legs, trunks and arms). Knee angle was estimated from the rowing machine seat position, measured with a linear encoder and transmitted wirelessly to a computer. Key results indicate the root mean square error (RMSE) between estimated and measured angles did not depend on group and stroke rate ($p>0.267$). Significantly greater RMSE values were observed however within the rowing cycle ($p<0.001$), reaching on average 8deg in the mid-recovery phase. Differences between estimated and measured knee angle values resulted in slightly earlier (5%) detection of knee flexion, regardless of the group and stroke rate considered. Offset of knee extension, knee angle at catch and range of knee motion were identified equally well with our model and with inertial sensors. These results suggest our model describes accurately the movement of knee joint during indoor rowing.**

***Index Terms*— knee joint, rowing, sensors, modelling.**

I. Introduction

Combining electrical stimulation with voluntary muscle activation has led to the emergence of promising paradigms for the cardiovascular and musculoskeletal conditioning of paraplegic persons [1], [2], [3], [4]. Of particular interest is the possibility of electrically eliciting knee extensors and flexors muscles while paraplegic subjects exercise on an indoor rowing machine: the FES Rowing [5], [6]. Although FES Rowing has been observed to outperform other exercise protocols [3], [5], [7], issues with the stimulation protocol remain open [8]. For example, fatigue of electrically stimulated muscles was reported to affect the training volume and thus the potential benefits of FES Rowing [1], [9]. Moreover, current protocols for FES Rowing unlikely reproduce lower-limb kinematics as smooth as those observed in able-bodied rowers [10]. The optimization of stimulation protocols during FES Rowing is therefore an issue of crucial relevance in paraplegia. In this view, patterning the stimulation profile according to the knee angle [11]–[13] likely constitutes a means to ensure muscles endure the electrically-assisted exercise for longer periods and thus to maximise the benefits of FES Rowing on health. For example, by monitoring knee joint angle, stimulation profiles could be shaped according to the velocity- and length-tension relationships of different muscles, ensuring stimulation intensity for each muscle is greater when its potential to produce force is maximal [14]–[16]. Shaping stimulation parameters demands, however, the monitoring of changes in knee joint during rowing.

Even though changes in joint angles have been measured during indoor rowing with both inertial sensors and video cameras [10], [17], [18], these measurement systems hardly suffice for the daily-use demands of FES Rowing. The time spent with subject preparation, system calibration and the cost and space associated with the use of conventional systems for motion analysis likely hinder the application of FES Rowing. An alternative practical solution would be the instrumentation of rowing machines with inexpensive sensors, which measurements could be regarded to identify the phases of the rowing cycle. Previous studies have indeed adapted

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commercially available rowing machines to provide real-time data on the position of the seat and of the handle though not on the joint kinematics [6], [7], [9], [10], [19]. Although the seat position has been acknowledged to fully define the configuration of the whole lower extremities during rowing [19], no evidence was found so far on the possibility of estimating knee joint angle from rowing machine data.

In this study, we propose a simple and affordable solution to estimate changes in knee joint angle during indoor rowing. First, we describe the adaptation of a rowing machine to provide real-time data on handle and seat position. Then, we present a subject-specific biomechanical model relating the seat position to the knee joint angle. Finally, changes in knee joint angle estimated with the adapted rowing machine are compared with actual changes in knee angle measured with inertial measurement units from a sample of elite rowers.

II. Methods

A. Adaptation of the rowing machine

We instrumented a commercially available rowing machine (Concept II model E, Morrisville, USA) to measure the position of the seat.

Seat position was measured with a custom-made linear encoder. The linear encoder consists of: i) a graduated adhesive bar placed along the surface of the rowing machine rail. Two 5 mm wide columns are printed on the top of the adhesive bar, each with alternating black and white marks (5.0×5.0 mm); ii) two couples of infrared emitter and receiver (QRE1113GR, Fairchild, USA). These sensors were mounted at the bottom rear of the seat, 2 mm distant from the rail. When the seat moves, the infrared beam is reflected alternately over the black-white-marking pattern and received by the photo-transistors, thus producing a continuous analog signal between 0V (black marks) and 3.3V (white marks). Transitions between voltage levels indicate a 5 mm shift of the seat on the rail. Movement direction was identified by comparing signals provided by the two couples of sensors, one for each column of black-white marks placed along the rowing machine rail. This measurement system ensured a resolution better than 10 mm, without hindering movement and without using cables.

B. Transmission of rowing machine data

Digital data from encoders mounted on the rowing machine handle and seat were collected synchronously with a custom-made system.

Position data was recorded with two wireless modules, one for the handle and one for the seat. Each module has a 16-bits ultra-low-power microcontroller (MSP430F5438A, Texas Instruments, Texas (USA)) that samples the position signals at 2048 Hz. Data is then sent to a Bluetooth transceiver through a standard Host Controller Interface (HCI) protocol implemented on a UART peripheral. Modules are powered by a 3.7V, 470 mAh 1-cell Lithium Polymer Battery (LiPo). A common pulse signal was sent to the microcontroller to trigger the start of data sampling after the connection of each module to the computer (receiver), resulting in a synchronization delay within ±500μs.

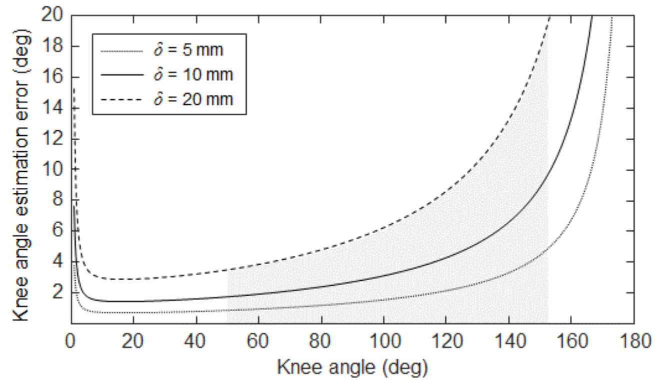

Fig. 2. Theoretical dependence of estimation error on the actual knee joint angle. Curves were computed from (5), by considering three systematic error values for the identification of joint rotation axes: δ = 5 mm (dotted line), δ = 10 mm (continuous line) and δ = 20 mm (dashed line). Shaded area indicates the minimal and maximal mean knee angle values observed here (cf. Results)

C. Biomechanical model: predicting knee joint angle from seat position

Here we describe the biomechanical model considered to estimate the knee joint angle from the rowing machine data. First, we defined the origin (O) of the Cartesian system of coordinates on the longitudinal axis of the monorail surface, at the maximal, forward seat position (Fig. 1a). Then, with respect to O, we measured the coordinates of the following four points on the flexfoot – the foot stretcher support where the foot is placed: the centre of the front (F) and rear (R) ends of the flexfoot and the medial (M) and lateral (L) edges of the rear border (Fig. 1a). Position F is the assumed point where force is applied to the stretcher and, when preparing for the catch, a degree of foot rotation around the methatarso-phalangeal area may occur (Fig. 1a); the model currently neglects this rotation. These coordinates were measured with the flexfoot at its maximal rear position. The vectors **v** and **f**, respectively $\overrightarrow{\mathrm{RF}}$ and $\overrightarrow{\mathrm{ML}}$, define the flexfoot plane. From **f** and **v**, we quantified the ankle joint position (A) with respect to R (Fig. 1b):

$$\overrightarrow{\mathrm{RA}} = a_h \frac{\mathbf{f} \times \mathbf{v}}{\|\mathbf{f} \times \mathbf{v}\|} + a_c \frac{\mathbf{v}}{\|\mathbf{v}\|} \tag{1}$$

Where × stands for the cross product. a_h and a_c respectively denote the height of the ankle from the ground and the distance between the ground projections of the ankle and of the tip of calcaneus, measured with the subject in the standing position. Ankle height was the distance from the tip of the lateral malleolus to the ground [20]. The ankle position in the coordinate system was therefore given by:

$$\overrightarrow{O\mathrm{A}} = \overrightarrow{O\mathrm{R}} + \overrightarrow{\mathrm{RA}} + (r-1)d \frac{\mathbf{v}}{\|\mathbf{v}\|} \tag{2}$$

With d indicating the distance (1.5 cm) between consecutive holes in the flexfoot along the **v** direction. The position of the flexfoot is given by the hole r at which the flexfoot is secured to the foot stretcher, ranging from 1 (flexfoot at maximal forward position) to 7 (Fig. 1).

After computing ankle position, we calculated the position of the hip joint (H) in relation to the ankle: $\overrightarrow{\mathrm{AH}} = \overrightarrow{O\mathrm{H}} - \overrightarrow{O\mathrm{A}}$. The coordinates of the hip joint in the x, y and z directions were respectively given by: the seat center position along the monorail, measured with the linear encoder (section II.A), half of the distance between the anterior superior iliac spines [20] and 4 cm above the seat surface.

Based on the law of cosines, the knee angle was:

$$\varphi = \cos^{-1}(\Delta) \tag{3}$$

$$\Delta = \frac{l^2 + t^2 - h^2}{2lt} \tag{4}$$

Where l, t and h respectively correspond to the length of the leg ($\|\overrightarrow{AK}\|$) and thigh ($\|\overrightarrow{KH}\|$) and the ankle-hip distance ($\|\overrightarrow{AH}\|$). The leg length was computed as the distance between the lateral malleolus and the popliteal fossa whereas thigh length was calculated as the distance between the popliteal fossa and the greater trochanter. It should be noted the proposed model is simple and may be implemented easily, given that: i) it applies to any rowing machine, providing the coordinates defining the foot stretcher surface and the seat center position are measured; ii) the anthropometric measurements can be undertaken at any time; iii) it demands the straightforward measurement of the flexfoot position used during rowing sessions; iv) it is not affected by hip movements outside the sagittal plane as, for a given seat position, the knee angle is the same regardless of whether a certain degree hip abduction/adduction occurs.

The definition of the joint axis of rotation according to anatomical landmarks certainly introduces a systematic error. Systematic errors associated with the measurement of leg (δ_l) and thigh (δ_t) length and of ankle-hip (δ_h) distance are presumably uncorrelated and lead to an estimation error for the knee angle (δ_φ) according to [21]:

$$\delta_\varphi = \sqrt{\left(\frac{\partial\varphi}{\partial l}\delta_l\right)^2 + \left(\frac{\partial\varphi}{\partial t}\delta_t\right)^2 + \left(\frac{\partial\varphi}{\partial h}\delta_h\right)^2} \tag{5}$$

$$\frac{\partial\varphi}{\partial l} = \frac{1}{\sqrt{1-\Delta^2}}\left(\frac{1}{t} - \frac{\Delta}{l}\right) \tag{6}$$

$$\frac{\partial\varphi}{\partial t} = \frac{1}{\sqrt{1-\Delta^2}}\left(\frac{1}{l} - \frac{\Delta}{t}\right) \tag{7}$$

$$\frac{\partial\varphi}{\partial h} = \frac{1}{\sqrt{1-\Delta^2}}\left(\frac{h}{lt}\right) \tag{8}$$

Two remarks follow from (5)-(8). First, δ_φ changes with variations in h. Second, δ_φ increases markedly as Δ approaches unity, which verifies whenever $h=|l-t|$ and $h=l+t$; i.e., whenever the knee angle approaches full flexion-extension. In absence of evidence on the actual value for the systematic errors in the identification of joint rotation axis, we assumed $\delta_l = \delta_t = \delta_h = \delta < 20$ mm. The dependence of δ_φ on h, and thus on φ, is shown in Fig. 2. Curves relating δ_φ to φ were created for three values of δ (5, 10 and 20 mm) and by considering the average values of l (429 ± 45 mm; mean ± st. dev.) and t (412 ± 39 mm) across all subjects tested (next section).

D. *Testing the biomechanical model*

Our model was tested on a sample of 26 rowers, split into two groups: 15 able-bodied elite rowers, with experience in international rowing competitions (range; 20-35 yrs; 55-86 kg; 160-192 cm), and 11 Paralympic-LTA rowers competing on the Rio 2016 Paralympic Games (23-47 yrs; 50-80 kg; 160-182 cm). All Paralympic athletes presented gentle motor impediments, not sufficient though to preclude execution of the complete rowing gesture: cyclic flexion and extension of legs, trunk and arms (LTA-PD Paralympic classification [22]). The inclusion of Paralympic athletes was expected to provide a more representative space of possible knee joint angle values for which we could test our model. Participants were recruited after providing written, informed consent and were informed on the experimental procedures, which conformed to the Declaration of Helsinki and were approved by the Ethic Committee of the Rio de Janeiro University Hospital (HUCFF-UFRJ; 57629016.7.0000.5257).

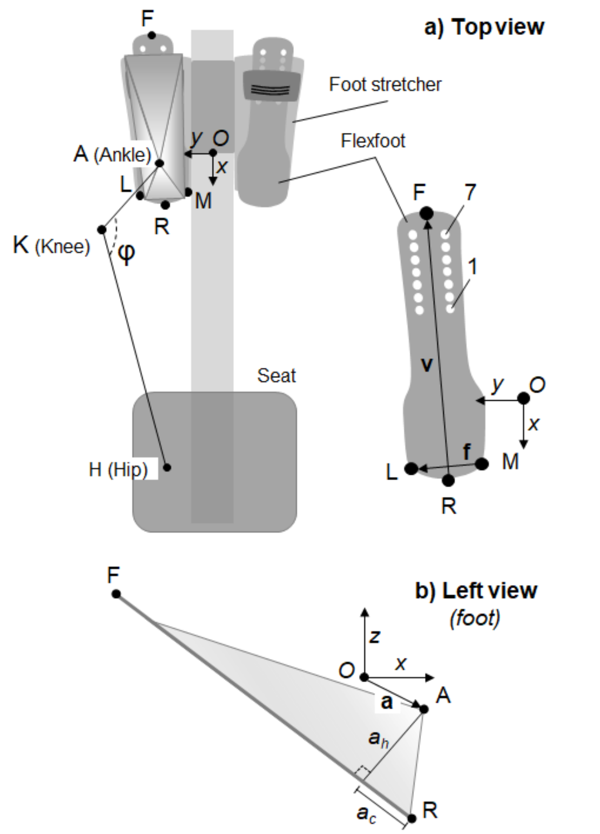

Fig. 1. Schematic representation of the rowing machine, viewed from top (a), and of the left foot position on the foot stretcher, viewed from the left (b).

After warm-up, participants performed three series of 30 strokes, each at a different stroke rate: 18, 24 and 32 spm. Participants were asked to perform as if engaged in a regular training session, with the rowing machine resistance and flexfoot position set according to their preference during all trials. Rest periods of at least 5 min were applied between trials.

The knee joint angle was calculated from the orientation of the leg and thigh segments. Orientation data was measured with two inertial sensors (MTx, XSens, Enschede, the Netherlands; angular resolution 0.05°; static accuracy (roll/pitch) < 0.5°; static accuracy (heading) < 1.0°; dynamic accuracy 2° RMS), tightly secured centrally and laterally to the left leg and thigh [23]. For each subject, sensors' positioning on thigh and leg was calibrated with the subject resting on the rowing machine with the knee at maximal extension; leg and thigh were considered to be located within a parasagittal plane. The experimenter ensured there was no relative movement between sensors and segments during acquisitions. Knee angle was then computed as in (3). It should be noted that, given an average velocity of the inertial sensors in the quasi-static range during experiments, the error of the knee angle values measured by means of inertial sensors is estimated to be lower than 0.5° [24]. Angle data obtained from the inertial sensors were, therefore, regarded as measured reference angles, against which we tested our model.

An additional inertial sensor was tightly secured to the machine handle, providing acceleration data that we used to identify individual rowing cycles according to [11]. Briefly, catch and finish instants were defined as acceleration peaks, identified from zero crossing instants in handle velocity. Catch denotes the onset of the drive phase, when rowers pull the handle towards themselves by extending their knees, hips, and shoulders and then flexing their elbows. Finish corresponds to the onset of the recovery phase, when rowers move forward in preparation for the drive. Acceleration and orientation data were sampled at 50 Hz (output frequency) with a 12 bit A/D converter and offline synchronised with the rowing machine data, via an external trigger pulse (TTL signal).

Differences between estimated and measured knee joint angles were assessed with the Root Mean Square Error (RMSE). First, measured and estimated knee angle data were segmented into 30 rowing cycles. RMSE values were then computed separately for each time instant, rowing cycle, group and stroke rate. After that, RMSE values were averaged over epochs corresponding to 10% of the rowing cycle, providing a total of 30 RMSE values per subject (3 stroke rates x 10 epochs). Rowing-related variables were computed to assess the relevance of RMSE values. From measured and estimated knee angle data we specifically computed the onset of knee flexion, the offset of knee extension, the knee angle at catch and the range of knee motion. Onset and offset values were computed from relative instants within the rowing cycle, when knee flexion respectively reached 10% of the knee range of motion during recovery and drive.

After ensuring the distribution of all data was Gaussian (Shapiro-Wilk statistics; $p>0.236$ for all cases) and the homogeneity of variance (Levene's test: $p>0.311$), parametric inferential tests were applied. One-way ANOVA was applied to assess differences in the duration of the rowing cycle and of the drive and recovery phases between groups (Able-bodied and Paralympic-LTA rowers), with stroke rate (18, 24 and 32 spm) as repeated measure. Two-way ANOVA was applied to assess the effect of group and stroke rate on RMSE values, with epochs within the rowing cycle (n=10) taken as repeated measure. The same statistical arrangement was considered to assess differences in rowing-related variables, with the procedure for computation of knee angle (estimated and measured) considered as repeated measure.

III. Results

All participants rowed at the requested stroke rate (Fig. 3a). No difference was observed in the mean duration of rowing cycles between groups, regardless of the stroke rate considered (ANOVA $F=0.04$; $p=0.837$). Differences between groups were however observed when considering rowing phases (group and stroke rate interaction; $F>4.081$; $p<0.03$ for both phases). Although the duration of recovery and drive phases decreased for both groups as stroke rate increased, at 18 spm the able-bodied rowers spent respectively more and less time than

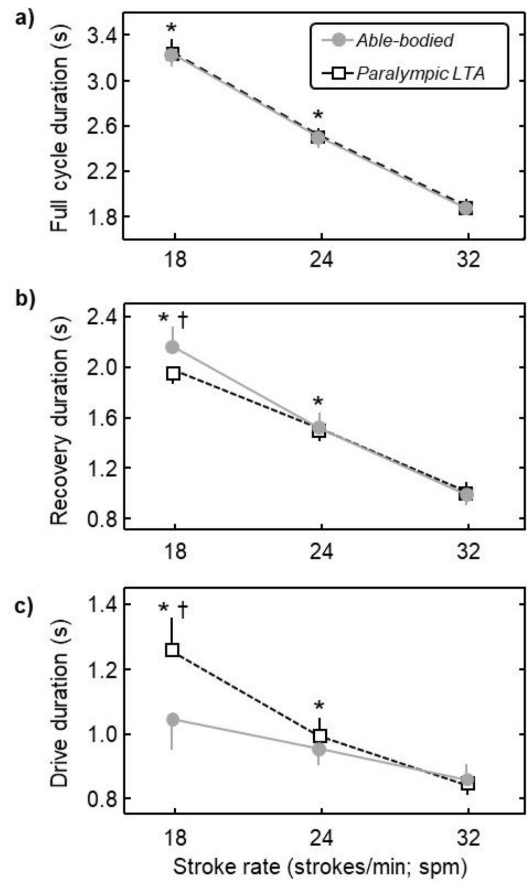

Fig. 3. Mean and standard deviation values (whiskers) computed for the duration of: the full rowing cycle (a), recovery (b) and drive (c) phases. Values are shown separately for the three stroke rates and for Able-bodied (●) and Paralympic-LTA rowers (□). * and † respectively denote significant differences w.r.t. greater stroke rates ($p<0.001$) and between groups ($p<0.002$).

Paralympic-LTA rowers in the recovery and drive phases (Fig. 3b and 3c; Bonferroni post-hoc analysis; $p<0.008$).

A. *Estimated versus measured knee joint angle*

Both group and stroke rate did not affect the estimation of knee joint angle. RMSE values computed between estimated and measured knee joint angle ranged on average from 3.8 to 5.1 deg across the three stroke rates (18, 24 and 32 spm) and the two groups (Paralympic-LTA and able-bodied rowers; Fig. 4). No interaction effect (group and stroke rate; ANOVA $F=1.102$; $p=0.342$) as well as no main effect was observed for group ($F=1.27$; $p=0.267$) and for stroke rate ($F=0.44$; $p=0.644$).

When considering different epochs within the rowing cycle, differences were observed between estimated and measured angles (main effect; ANOVA $F=31.199$; $p<0.001$). Inspection of knee angle profiles computed from the rowing machine and from the inertial sensors data for an able-bodied participant, representative for both groups, reveals greatest differences when the knee reaches maximal extension and on the initiation of knee flexion (Fig. 5a). Paired analysis on all participants of both groups indicates the RMSE values were significantly greater during 20%-50% and 80%-100% of the rowing cycle, roughly reaching 8 deg on average during mid-recovery (Bonferroni post-hoc analysis; $p<0.003$ for all cases; Fig.5b). Post-hoc analysis did not reveal any difference between groups ($p>0.4$ for all cases), even though differences in RMSE values between able-bodied and Paralympic-LTA rowers tended to increase towards the end of the rowing cycle (Fig. 5b).

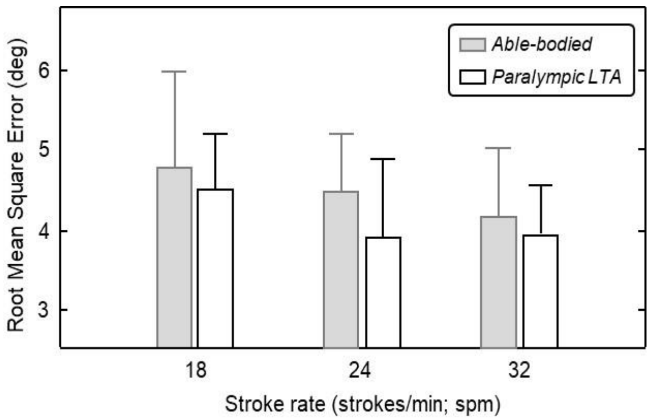

Fig. 4. Mean and standard deviation (whiskers) values computed for the Root Mean Square Error between estimated and measured knee angle. Grey and white bars respectively denote Able-bodied and Paralympic-LTA rowers.

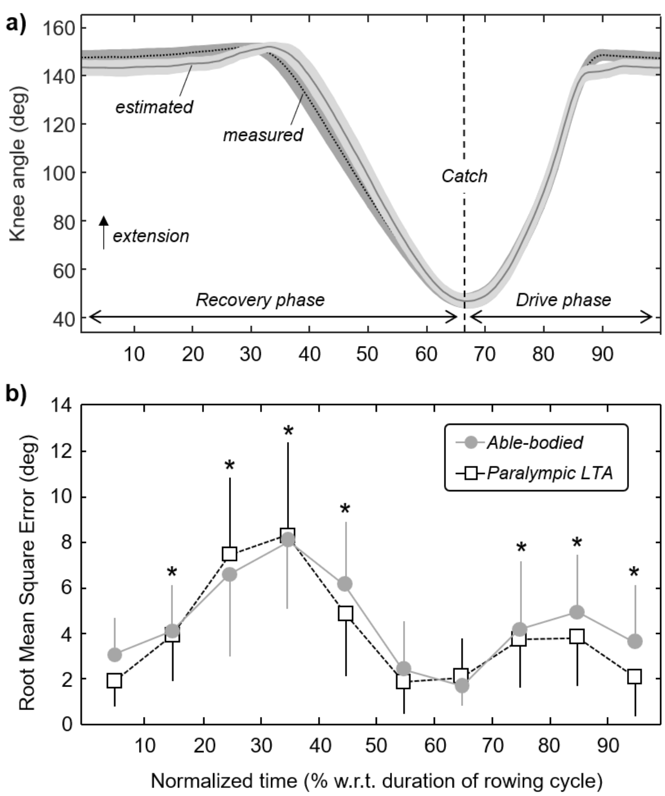

Fig. 5. Knee joint angle profiles averaged across 30 rowing cycles are shown in a) for a single, Able-bodied rower. Continuous and dotted lines indicate mean angle values whereas the standard deviation is indicated with shaded areas. The vertical, dashed line denotes the average, catch instant. Mean and standard deviation (whiskers) for RMSE values are shown in b) for both groups and separately for each of the 10 epochs of the rowing cycle. * indicates significant differences w.r.t. the smallest, mean RMSE value across groups (1.9 deg at 70% of the rowing cycle; $p<0.003$).

B. Timing and amplitude of knee movement

Differences in the onset of knee flexion were observed between estimated and measured knee angle values. Onset values computed for knee angle estimated from the rowing machine were consistently greater by roughly 5% than those obtained from measured angle values, regardless of the group and stroke rate considered (Fig. 6a; main effect ANOVA $F=121.321$; $p<0.001$). For both estimated and measured knee angle values, Paralympic-LTA rowers exhibited early onset of knee flexion than Able-bodied rowers when performing at 18 and 24 spm (Fig. 6a; main effect between groups; $F=14.118$; $p<0.001$). No interaction effect between the procedure for knee angle computation and stroke rate or group was observed ($F<0.591$; $p>0.520$). Similarly, no significant differences in the offset of knee extension, knee joint angle at catch and range of knee motion between estimated and measured knee angle values were observed for both groups and for the three stroke rates tested (Fig. 6b-d).

Values for angle at catch and for the range of motion define the absolute excursion for the knee joint during rowing; subjects moved their knee from 48 ± 9 deg to 152 ± 9 deg on average (mean ± std. dev.; N=26 subjects). These values define the operating range over which the effect of uncertainties in the definition of joint rotation axes on the estimation of knee angle may be assessed (cf. shaded area in Fig. 2).

IV. Discussion

In this study we present a subject-specific biomechanical model to estimate knee joint angle from a modified rowing machine. From data collected from a sample of elite, Able-bodied and Paralympic-LTA rowers, we asked two questions: 1) how accurately knee joint angle may be estimated from changes in the seat position; 2) whether differences between estimated and measured knee angle may lead to appreciable differences in rowing-related variables.

A. Able-bodied versus Paralympic-LTA rowers

By testing our model on a sample of able-bodied and Paralympic-LTA we expected to cover a wide range of knee joint angle profiles than we would have done by testing either group. While previous studies reported high intra- and inter-individual consistency during indoor rowing for able-bodied elite rowers [17], [23], [25], [26], similar data on Paralympic-LTA rowers are missing. Although interest on para-rowing kinematics seems to be emerging [27], to our knowledge this is the first study reporting kinematics data for a sample of Paralympic-LTA rowers. Even though the duration of rowing cycles increased similarly in both groups as stroke rate decreased (Fig. 3a), Paralympic-LTA rowers spent respectively less and more time in the drive and recovery phases when performing at lower stroke rates (Fig. 3b,c). This likely explains why Paralympic-LTA rowers flexed their knee earlier than able-bodied rowers did, in particular for lower stroke rates (Fig. 6a). Notwithstanding these group differences in indoor

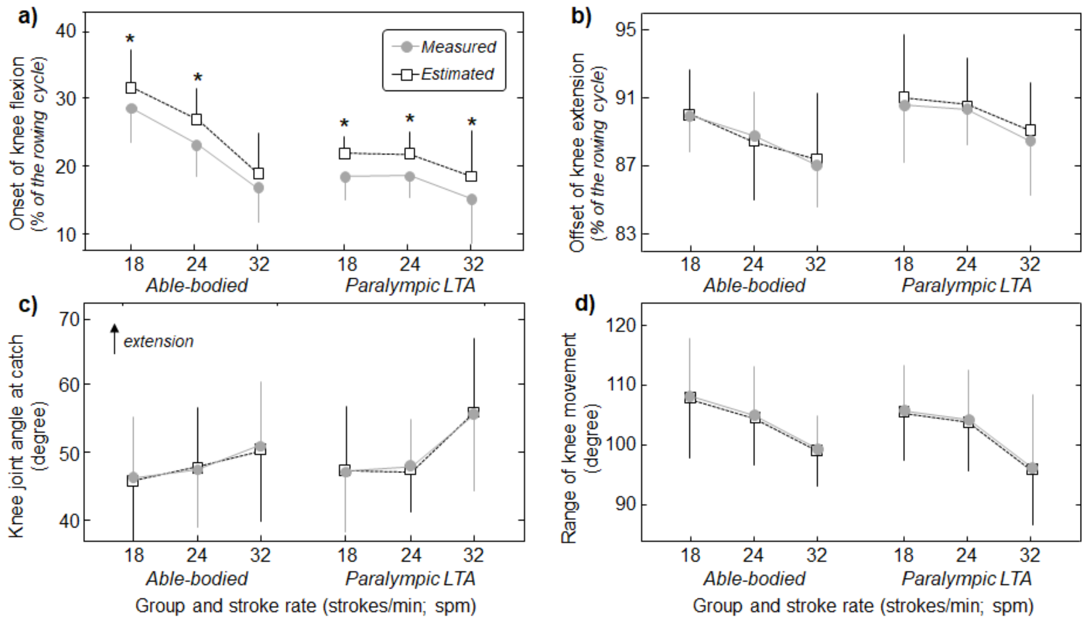

Fig. 6. Mean and standard deviation values (whiskers) computed for the onset of knee flexion (a), offset of knee extension (b), knee angle at catch (c) and range of motion of the knee joint (d). Data are shown separately for the three stroke rates, for Able-bodied (left) and Paralympic LTA (right) rowers and for measured (●) and estimated (□), angle values. * denotes significant differences between estimated and measured values (p<0.001).

rowing performance, our model provided similar estimates of changes in knee joint angle for both groups.

B. Differences between measured and estimated knee joint angle during rowing

On average, differences between estimated and measured knee joint angle amounted to 5 deg. This figure represents roughly 4% of the range of knee motion observed during rowing (Fig. 6d; see also [10], [23], [28]). Even though the average RMSE values did not depend on group and stroke rate (Fig. 4), the estimation error differed significantly within the rowing cycle. Greater RMSE values were observed during mid recovery, after knee full extension, and on the initiation of knee flexion (Fig. 5b); knee flexion angle was underestimated when the seat was not moving and overestimated on the onset of knee flexion (Fig. 5a). Such discrepancies were likely due to the relative movement between the hip joint and the seat. It is common experience during rowing that, during the drive phase, the knees are moved vigorously into extension. Given the foot stretcher and the seat are at different heights, with the former being at a lower position, vertical forces on the seat approach 0 N at the end of knee extension [28]. Considering knees often move into hyperextension at the end of the drive phase and the trunk moves into extension, it is thus possible that maximal knee extension has been reached after the seat ceased moving from mid-drive to mid-recovery phase (Fig. 5a). Similarly, the hip movement in relation to the seat may explain the significant differences between estimated and measured angle values at the onset of knee flexion (Fig. 5). Interestingly, immediately prior to knee flexion, estimated values suggest there is a small degree of knee extension (cf. estimated knee angle during ~25%–35% of the rowing cycle in Fig. 5a). Advancing any possible mechanisms accounting for this discrepancy may be currently speculative, as we did not quantify motion other than that of the knee joint. Nevertheless, it should be noted the trunk approaches its maximally flexed position at mid-recovery [17], possibly moving the seat backward. Given our model (3) conceives seat backward movements as knee extension, any seat motion induced by e.g. trunk movements would be interpreted incorrectly as knee motion. Regardless of the actual sources of discrepancy between estimated and measured values, current results indicate differences may exist between changes in seat position and knee angle. Different from previous claims [19], measuring seat position seems not equivalent to measuring knee joint angle.

In spite of the significant differences between estimated and measured knee angle values, our biomechanical model provided accurate estimates for most of the rowing-related variables assessed here. Except for the onset of knee flexion (Fig. 6a), the offset of knee extension, the knee angle at catch and the range of knee motion were estimated remarkably well (Fig. 6b-d). The onset of knee flexion was consistently underestimated by roughly 5% for the two groups and the three stroke rates, suggesting there was a systematic, estimation error. Systematic errors affecting the estimation of knee flexion onset were more likely due to relative movements between joints and seat, as just discussed, than to uncertainties in the definition of joint rotation axes from anatomical landmarks. On one hand, Fig.2 indicates large uncertainties may lead to estimation errors as large as 20 deg. On the other hand, these uncertainties were sufficiently small (<10 mm), otherwise discrepancies between measured and estimated knee angle smaller than 10 deg (Fig. 5b) would be unlikely appreciated. Moreover, if errors in the estimation of knee angle were mostly due to the definition of joint rotation axes, we would expect greatest discrepancies between estimated and measured knee angle values when knee was at maximal extension (Fig. 2). Results shown in Fig. 5 suggest this was not the case however; RMSE values were not constant when the knee was kept at maximal extension, from mid drive to mid recovery. The physiological or functional significance of estimation errors affecting our model depends on the application at which its use is aimed. For example, the 5% difference observed here (Fig. 6a) is smaller than the inter-individual variability reported for the onset of activation of lower limb muscles in able-bodied rowers [10], [18], suggesting the estimation error resulting from our model is not crucial. It seems, therefore, the model presented here (Fig. 1; (3)) describes sufficiently the changes in knee joint angle occurring during indoor rowing, opening new fronts for applications focused on the stimulation of lower limb muscles in paraplegia (FES Rowing [2], [5], [14], [29]).

C. The potential relevance of estimating knee joint angle during rowing

It is the movement of the lower limbs, and thus the change in knee joint angle, that mostly determines rowing performance. Indeed, the success of FES-Rowing protocols consists in the cyclic and timely stimulation of lower limb muscles, actively guiding paraplegic subjects through the rowing recovery and drive phases. Early FES-Rowing protocols were based on the manual determination of stimulation instants, using either constant or variable stimulation intensities [5]–[7], [10]. Advanced FES-Rowing systems have been developed, relying on the seat kinematics to automatically pattern the electrical stimulation of knee flexors and extensors [14], [19]. Two issues common to both types of FES-Rowing systems are the inability to produce lower limb movements as smooth as those observed in able-bodied rowers [10], [14] and, most importantly, the early development of muscle fatigue [1], [14]. Muscle fatigue may limit the duration and exercise volume of a FES rowing session, which, in turn, would reduce the potential cardiovascular benefits in those with spinal cord injury. We believe the possibility of introducing real-time estimates of knee angle in FES-Rowing systems may assist in circumventing these issues. First because stimulation patterns could be issued to produce knee angle profiles similar to those observed for able-bodied rowers or according to individual needs. When attempting to control knee joint position through quadriceps stimulation, Chang et al [11] observed that the inclusion of a proportional-integral-derivative feedback controller resulted in smaller tracking errors between the actual and desired trajectories. Second, stimulation intensity could be modulated in accordance with the potential of knee muscles to produce force actively (i.e., based on the muscle velocity- and length-tension relationships). Even for a fixed degree of excitation, for example, it is well established that force changes with both the muscle length and shortening-lengthening velocity [14], [15]. The dependence of force production on the muscle intrinsic properties has been suggested to further vary between voluntary and electrically elicited contractions [16]. Stimulating knee muscles with a constant intensity or patterning stimulation without considering the muscle intrinsic properties may consequently lead to the early development of fatigue, in addition to producing movement patterns different from those expected during rowing. Patterning electrical stimulation based on knee muscle properties demands, however, the possibility of monitoring changes in knee joint angle during rowing.

Current results suggest the model presented here may constitute a feasible means to quantify knee joint angle during rowing. The main advantage of our model is the possibility of estimating knee angle without the need of: i) securing sensors or markers to the subjects' limbs; ii) using expensive and unwieldy systems, hindering the application of FES Rowing outside the laboratory setting. Anthropometric measurements necessary for implementing the model may be collected at any time from subjects. The only variable to consider in the model (2) is the flexfoot position on the foot stretcher, which may be measured prior to commencing the rowing session. In spite of the 5 deg difference between estimated and measured angle values (Fig. 4), which relevance in rowing applications (e.g. FES Rowing) remains to be determined, results from two samples of elite rowers with different characteristics (Fig. 3) suggest the knee joint angle may be estimated accurately during indoor rowing.

There are some issues that deserve further investigation prior to generalizing our model to different rowing machines and subjects. As currently conceived, the model could be readily implemented for different rowing machines, providing the coordinates defining the foot stretcher surface and the seat center position are measured. Concerning generalization for different populations, two issues should be considered. First, ankle coordinates in the spatial reference system may change depending on the individuals' footwear. These changes could be accounted for in the model by shifting a_c and a_h according to the thickness of the shoes-feet and shoes-rear flexfoot support interfaces respectively. Both could be measured readily at any time, even though they are likely shoes-specific. A second note concerns the foot rotation around the methatarso-phalangeal joints, which has been neglected in the present model. Although current results suggest this assumption may stand, at least for the sample studied, it is possible that estimation errors may be larger for subjects suffering from e.g. equinus contracture. At the moment we are unable to provide figures on how much the amount of foot rotation may affect the model. Nevertheless, it should be noted foot rotation verifies near catch position, where measurements errors in ankle joint position are relatively low.

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