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Development of a Device for Measurement of Multi-Directional Isometric Trunk Kinetics in a Seated Position

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Abstract

Trunk torque is typically quantified in a single plane. This is not ideal since any unmeasured coupled trunk kinetics in other directions, than the intended one, could affect the accuracy and reliability of the strength measurements as well as the ability to corroborate findings with electromyographic recordings. Therefore, an isometric device that simultaneously records trunk kinetics across planes has been developed to aid in the research of trunk control in both the healthy or impaired populations. This device utilizes a six degree-of-freedom load cell and custom designed frame to attach individuals while in the sitting position. The performance of the device was tested in six healthy controls and while using two protocols. The device was able to detect coupled trunk kinetics during maximum lateral flexion and axial twisting torque generation. It also allowed the implementation of a multi-axis isometric protocol showing that subjects were able to generate larger amounts of axial torque during sub-maximal trunk extension compared to sub-maximal trunk flexion. In conclusion, the device and mechanical transformations discussed in this article will aid in the interpretation of multi-directional isometric trunk kinetics in a wide range of populations.

Index Terms

Isometric; torque; trunk

I. Introduction

There is a general lack of multi-directional isometric trunk kinetics research related to musculoskeletal and neurological movement disorders [1]–[7]. Due to the inherent complexity of the spinal vertebral column and overlying trunk musculature, having the

ability to simultaneously obtain these multi-directional measures of trunk kinetics is difficult but necessary [1], [4], [5].

Trunk impairment has been typically quantified by the magnitude of trunk force or torque generated in a single direction by handheld force sensors and one degree-of-freedom (1-DOF) dynamometers which at most provide an estimate of strength during isometric and isokinetic trials [2], [3], [8]–[10] in a single anatomical plane. The shortcomings of these devices are that they do not provide any synergistic torque coupling of trunk kinetics that may be simultaneously occurring in directions other than the primary intended direction (e.g., unintentionally generating axial torque in the transverse plane or lateral torque in the coronal plane during intentional trunk flexion in the sagittal plane). Any unmeasured synergistic torque coupling in other directions than the intended one could affect the accuracy and reliability of the strength measurements as well as the ability to corroborate findings with electromyographic recordings [5]. More importantly, any presence of synergistic torque coupling in the trunk may be related to osseoligamentous constraints of the spine, compensatory strategies of the individual or abnormal muscle coactivation patterns resulting from neuromuscular pathologies [11].

Coupling of spine kinetics and kinematics has been previously observed both *in vitro* and *in vivo*. For example, coupling between lateral bending in the coronal plane and axial twisting in the transverse plane [12] have been demonstrated in cadaveric preparations as well as during dynamic rotations [13] and under isometric conditions in human participants [11]. It has been hypothesized that this coupling may be the inherent nature of the spine due to the passive ligamentous structures and the vertebral facets [14]. Alternatively, model simulations have suggested that passive structures cannot completely explain the nature of this coupling and that muscle co-activation patterns - must influence this behavior [15]. Since there is no single agonist with a muscle force vector that solely creates axial torque, it would seem plausible for multi-planar movements to be the natural result of muscle contraction during these movements [16]. Therefore, if muscle activation can influence the coupling of observed trunk kinetics, the coupling may differ based on the type of neuromuscular pathology. This has been shown for individuals with low back pain in that differences in synergistic torque coupling and muscle activation patterns under isometric conditions were different than those compared to healthy controls [11]. Thus, in the study of both healthy individuals and those that have neuromusculoskeletal pathologies, it is important to obtain multi-directional measures of torque and muscle activity to capture both in and out of plane trunk torques which will be referred to as coupled trunk kinetics (CTK).

There currently exist few devices available to measure CTK. The devices that do allow for 3-DOF isometric trunk torque measurements have known deficiencies for multi-plane kinetics due to the inherent difficulty of participant attachment and/or alignment with sensor axes [4]. These deficiencies are a direct consequence of the fact that joint axes and adequate rigid contact points are not explicitly defined for the trunk region while preserving a tolerable level of comfort for the participant. Another shortcoming is that most of the current devices require participants to involve the use of lower extremities by having to stand [6] or involve the use of upper extremities to operate the device correctly [17]. Such requirements may limit the types of cohorts that can participate if extremity use is compromised such as

observed following stroke, cerebral palsy, head trauma or multiple sclerosis. A device that can simultaneously record multi-directional trunk kinetics without the requirement of lower or upper extremity involvement may increase accuracy, repeatability and accessibility to many cohorts thereby improving estimates of trunk kinetics in both able-bodied participants and individuals with neuromusculoskeletal pathologies. The successful implementation of such measures are likely to increase our understanding of CTK that naturally exist in able-bodied individuals during the generation of isometric trunk torques. In an individual with pain, musculoskeletal or neurological dysfunction, CTK that deviates from what is expected could be related to abnormal sensorimotor control or fatigue of the trunk musculature [5]. A better understanding of CTK will provide an additional measure of identifying trunk impairment outside the typical single DOF strength measures reported previously [2]. In short by adding multi-DOF measurements, better characterization of normal and impaired trunk control is anticipated that may subsequently lead to the creation of novel treatment interventions.

The aim of this work was to present a device that fulfills the previously discussed design criteria in an effort to improve estimations of CTK across the three anatomical planes of trunk movement. Our main goals for the device were to simultaneously record kinetics in each anatomical plane while providing accessibility to a large range of cohorts by means of participant attachment and comfort.

II. Description of Methodology

A custom designed apparatus was developed to simultaneously collect isometric trunk kinetics across each anatomical plane (flexion/extension—sagittal plane; lateral bending—coronal plane; axial twist—transverse plane) while in the seated position. The outer structural frame of the apparatus contains a seat, lumbar support, lateral pelvis/hip restraints, and an attachment used to restrain the pelvis against the seat cushion (Fig. 1). The seat is adjustable in seat depth to allow accommodation of varying leg lengths between individuals. The seat is also adjustable such that the posterior portion of the seat tilts down (not shown) to shift pressure from the is-chial tuberosities to the thighs to better approximate the lumbar lordosis and sacral inclination typically seen in standing [18]. The lumbar support is a narrow curved piece of plastic with viscoelastic lining that is adjustable in height and depth so that it can be placed at the lumbar level of the participant's spine to support lordosis and prevent back injury. A 1-DOF load cell (M-31, 250 lb, Honeywell, Columbus, OH, USA) is placed in line with the lumbar support to measure resultant forces of the spine in the posterior direction (Fig. 1). The narrow design of the lumbar support allows little to no pressure on electrodes that may be used to simultaneously record muscle activity of the lumbar trunk extensors. The lateral pelvis restraints are cushions that are adjustable in depth so that the pelvis is restrained from lateral rotation and translation. The anterior pelvis restraints contacts the anterior portions of the pelvis on a downward angle with cushions that are adjustable in depth and angle in order to restrain the pelvis from flexing and extending relative to the seat. The inner frame consists of four cushions that are adjustable in depth and contact the anterior, posterior, and lateral portions of the individual's upper body (Fig. 1).

The inner and outer frames are connected to each other by a 6-DOF load cell (JR3 Inc., Woodland, CA, USA) so that any torque or force from the inner frame, relative to the outer frame, is transmitted through the 6-DOF load cell. The orientation of the 6-DOF load cell are such that recorded forces are positive in the left (Fx), superior (Fy), and anterior (Fz) directions and the moments along each axis are positive in the clock-wise direction (Fig. 2). The feet and lower legs of a seated participant do not contact the floor so that the only constraint to the lower extremities is the seat cushion that prevents any dynamic hip extension from occurring. In order to account for differences in postural adjustments that may occur during trunk torque generation, two 1-DOF load cells (S-Beam, 250 lb, Omegadyne, Sunbury, OH, USA) are placed under the distal portion of each thigh to measure any resultant forces that occur if the participant generates isometric hip extension against the load cell (Fig. 1).

The device is developed to ensure ease of getting into and out of it in a timely manner and in the case of emergency. In order to attach the participant to the device, the inner frame is rotated backwards via hinge joints in order to seat the participant [Fig. 3(A)]. Once the participant is seated, a shoulder strap restraint mechanism (shoulder restraint, <http://www.pxdirect.com/grip.htm>) is placed directly on the subject to attenuate shoulder movement relative to the trunk. Subsequently, the inner frame is rotated forward over the participant [Fig. 3(B)] and locked into place. The inner frame is then adjusted vertically so that the center of the 6-DOF load cell and vertical center of the inner frame were at the height of the T3 spinous process of the participant. This location was chosen as the highest level in which the lateral and anterior cushions of the inner frame could have the tightest attachment points with the shoulders and sternum, respectively, while also maintaining comfort to the participant. Subsequently, the anterior and lateral cushions on the inner frame are adjusted in depth to secure the participant and center their spine relative to the 6-DOF load cell. Once the participant's upper body is secured to the inner frame, the lower body is secured to the outer structural frame using the anterior and lateral pelvis restraints [Fig. 3(C) and (D)]. Lastly, the lumbar support is adjusted in depth to the participant's desired comfort. Prior to the start of the chosen experimental protocol, 3-D coordinates of targeted spinous processes, in relation to the 6-DOF load cell (dx, dy, and dz), can be obtained using motion tracking or digitizing software in order to resolve the kinetics at a predetermined location (Fig. 4).

III. Evaluation

A. Participants

Six healthy control subjects (4M/2F, 62.5 ± 6.7 years, 175.9 ± 8.6 cm, 74.5 ± 10 kg, 24.0 ± 2.1 BMI) participated in the device evaluation using both single-axis and multi-axis isometric protocols. The main exclusion criterion for control participants was any history of neurological disorder, any formally diagnosed chest-pain, low back pain or surgery, and a body mass index greater than 30.

B. Subject Attachment

The seat depth was adjusted so that each subject maintained a hip flexion angle of 90° . The 6-DOF load cell and inner frame were then adjusted vertically to be at the vertical height of the T3 spinous process of each subject. The subject's pelvis and trunk were then centered in relation to the 6-DOF cell using manual measurements of the lateral pelvis restraints on the outer frame and lateral trunk restraints on the inner frame. Next, location and orientation of the 6-DOF load sensor were defined in the global frame using a high-resolution 3-D digitizing device (Optotrak Certus Motion Capture System, Northern Digital, Inc., Waterloo, ON, Canada). Since the axis of rotation is thought to be near the third vertebra of the lumbar region (L3) for sagittal movements [19] without being explicitly defined for other planes, L3 was digitized on each participant in the global frame and later translated into the local coordinate frame of the 6-DOF load sensor. The measured location of L3, in relation to the 6-DOF coordinate frame, was used to resolve the forces and moments generated in the sagittal, coronal and transverse planes, respectively, to the L3 region using the translation matrix (Fig. 4).

C. Protocol

1. Participants were first asked to perform a single-axis isometric protocol by generating isometric torques only in a specified direction in a specified plane using 1-DOF feedback information along with verbal encouragement from the experimenter. More specifically, they were asked to perform maximum voluntary torques (MVTs) of gradually increasing isometric torque generation only in the sagittal (flexion and extension), coronal (left and right), and transverse (left and right axial twist) planes to the best of their ability resulting in a total of six directions that were randomized for each participant. Using a custom designed LabVIEW algorithm (LabVIEW v7.1, National Instruments, Austin, TX, USA), feedback of only torque in the instructed direction was provided and depicted by a bar graph that increased and decreased along with the level of torque being generated in the specified direction.
2. Subjects were then asked to perform a multi-axis isometric protocol by generating MVTs in the transverse plane while simultaneously targeting sub-maximal torque (25% or 50%) generation in the sagittal plane (flexion or extension). 2-DOF feedback was provided in vector form in that sagittal torque increased and decreased the length of a thick line on the screen while the amount of axial torque changed the angle of the line in the clockwise or counter-clockwise direction. The vector origin was fixed at the center of two concentric circles that represented the minimum and maximum allowable sagittal torque range. The radius of the outer circle representing the upper sagittal torque range (25% or 50% plus 5%) and the radius of the inner circle representing the lower sagittal torque range (25% or 50% minus 5%). Along with the visual feedback, a beeper sounded an audible tone when subjects were within the allowable torque range. Successful trials for both protocols were defined as having less than 10% difference between the maximum values obtained in three subsequent trials. If greater differences were obtained, additional trials were administered to get the "actual" MVT and this was verified during the

experiment. Two minutes of rest were given between each trial with each trial lasting from 2–6 s.

D. Analog Filtering and Data Collection

On-site analog filtering was performed on load cell signals prior to being sampled to prevent aliasing. Load cell signals were low-pass filtered at a cut-off frequency of 40 Hz. Filtered signals were sampled at a rate of 1000 Hz using a custom-designed data collection program in LabVIEW.

E. Data Postprocessing

All collected data was postprocessed and further analyzed using custom-designed software in Matlab (r2008a, The Math-Works, Inc., Natick, MA, USA). Load cell data were low pass filtered using a zero-phase eight pole Butterworth filter with a cut-off frequency of 5 Hz in order to eliminate noise.

For the single-axis protocol, torque data in the sagittal, coronal and transverse plane were divided into two categories: primary and secondary torques related to CTK. Primary torques were the torque direction in which the participant was instructed to perform their MVT using the 1-DOF feedback. The remaining two directions of torque for that trial are referred to as secondary torques. The criteria for primary torque detection were defined as the largest torque value that did not vary by 10% for a minimum of 500 ms. In this way, inadvertent spikes or jerky motions were not regarded as maximum values. The time index of the maximum primary torque was used to obtain the secondary torque values. Averages of the primary and secondary torques were then taken across the three successful trials and used for further analysis. For the multi-axis protocol, only transverse torque that occurred while the subject maintained sagittal torque in the allowable torque range were inspected using the same criteria already described. The maximum transverse torque was then normalized to the highest transverse torque that each subject was capable of generating in the single-axis protocol.

F. Outcomes

1) Single-Axis Isometric Protocol—Maximum primary and corresponding secondary torques are reported for each of the six directions in which subjects were asked to generate MVTs in the single-axis isometric protocol.

2) Multi-Axis Isometric Protocol—Maximum transverse torques that subjects could generate during the 2-DOF visual feedback protocol are normalized to the maximum transverse torque the respective subjects were able to generate during the single-axis isometric protocol.

3) Sensitivity Analysis—In order to determine the effect of measurement error by the experimenter and/or movement of the subject on the translation of recorded kinetics output of the 6-DOF load cell, a sensitivity analysis was performed on data obtained during performance of the single- and multi-axis isometric protocols. This was done by using the original translated kinetics output from each individual and comparing them to the translated

kinetics outputs when a combined error of 10 mm was added to each axis (i.e., x, y, and z) of the original translation distance. The resulting absolute difference on each torque axis of the kinetics values, before and after the addition of error, was then normalized to the maximum torque the subject generated for that axis. The group medians and ranges of normalized error for each primary direction are reported.

IV. Evaluation Results

A. Single-Axis Isometric Protocol [Fig. 5(a)]

Subjects generated little to no secondary torques when attempting to generate maximum torques in the sagittal plane. Subjects did generate secondary torques when attempting to generate maximum torques in the coronal plane in that they coupled extension torque with primary lateral flexion to the right and left side. Similar results were seen during axial twist in the transverse plane in that subjects coupled extension torque with axial twist, with larger extension torque occurring when twisting towards the left side.

B. Multi-Axis Isometric Protocol [Fig. 5(b)]

When attempting to generate maximum axial torque with trunk flexion, subjects were able to generate less axial torque than in the single-axis isometric protocol as well as when coupling axial torque with trunk extension.

C. Sensitivity Analysis (Fig. 6)

During the generation of maximum torque in all six directions (Fig. 6(a); single-axis isometric protocol), errors as a percentage of maximum torque in the sagittal plane ranged from 0.0% to 5.2%. In the coronal plane errors ranged from 0.2%–7.6% and in the transverse plane errors ranged from 0.5%–14.9%. During the generation of maximum transverse torque while holding sub-max torques in the sagittal plane (Fig. 6(b); multi-axis isometric protocol), errors as a percentage of maximum torque in the sagittal plane ranged from 0.2%–4.2%. In the coronal plane errors ranged from 0.0%–3.4% and in the transverse plane errors ranged from 0.2% to 9.8%.

V. Discussion

The aim of this article is to discuss the development and testing of a novel device with flexibility in terms of single and multi-degree of freedom trunk torque estimations while also broadening the types of pathologies that can be studied. In summary, our approach involves the development of a custom frame and the utilization of virtually translating a 6-DOF load cell to a specified location. Our novel use of a 6-DOF load cell to measure trunk kinetics prevents errors that can occur from having sensor axes that are fixed in a global frame relative to the subject. Another benefit of using a single 6-DOF load cell, compared to devices that typically use three 1-DOF torque sensors for each anatomical plane, is that torque cross-talk resulting from misalignment of sensors is inherently decreased (JR3 Inc. 6-DOF load cells have 0.0% of cross-talk via 6×6 calibration matrix). Furthermore, using a 6-DOF load cell resolves all kinetics being generated wherein 1-DOF torque sensors can only measure perpendicular forces being applied to the attached lever arm. Considering the

degrees of flexibility in the spine even under the best constrained conditions, it would seem very likely that loads are not always placed perpendicular to the lever arm and would cause kinetics information to be lost. Furthermore, considering the lack of joint axis definitions for each plane of the total spine motion, our device allows for digitizing multiple regions of the spine that are plane specific as that data becomes further defined. Another unique aspect of our device is the allocation of 1-DOF load cells behind the lumbar support and under each leg. These sensors add quantitative information regarding the postural adjustments being used by the participant during the generation of max and sub-max trunk torques. Table I shows data from the 1-DOF lumbar and each respective leg load cell from a single subject during the single-axis isometric protocol. Values were obtained using the time-index of the maximum primary torque for each direction. For this subject, it can be seen that during trunk flexion, this subject generated posterior loads on the lumbar load cell and no downward force on each of the leg load cells. Alternatively, during trunk extension, there was no load placed on the lumbar load cell, but downward loads from each leg were generated. During lateral flexion to the left and right, the leg opposite to the side the subject was laterally flexing towards placed downward loads on the respective load cell. During axial twist, loads were placed on both load cells simultaneously with greater loads being placed on the same leg as the side they were axially twisting towards. Considering the difficulty in adequately constraining the pelvis and trunk, having access to this information allows the experimenter to adjust the level in which they want to control the participants' outer frame contact load through the use of audio or visual feedback. It may at the least provide insight when participant data seems unique of the population being studied. Furthermore, comparison of the spontaneous postural loads between cohorts can further validate observed differences from trunk kinetics if no significant difference in postural loads is found. Considering the location of the trunk region in the kinematic chain of the whole body, this is especially important for pathologies such as stroke where abnormal synergies of upper [20] and lower [21] extremities are present during maximum exertions and could accompany max exertion of the trunk. Another benefit of our design is that the custom designed frame also allows participants to generate trunk torques while in a seated position which increases comfort, decreases fatigue for extended studies and increases the range neuromuscular pathologies that can be researched.

It is important to note possible design limitations of our device. One limitation is the lateral attachment method used on the inner frame of the current device. In order to best couple the participant to the load cell, we applied pressure on four sides of the participant as opposed to strap methods typically used in other multi-directional setups [4]. Constraints are placed external to the shoulder joint to better couple the participant to the device and so that the musculature of the shoulder joint would help dissipate sustained loads as opposed to direct contact on the skin and lateral bones of the rib-cage which may cause discomfort. Subsequently, since the device constraints are placed external to the shoulder joint there is a chance that asymmetrical shoulder flexion/extension and abduction can be measured by the 6-DOF load cell and create loads that do not necessarily originate from the axial trunk musculature. To address this, we applied the shoulder strap restraint mechanism previously described to attenuate all shoulder movements relative to the torso and thus minimize the magnitude to which this could occur. We believe that the type of shoulder restraint to be

used should not be too large or cumbersome as that may affect the mobility of the spine and ability of the latissimus dorsi muscle to transmit loads to the 6-DOF load cell since it is involved in extension and axial rotation of the trunk. Another limitation worth noting is the fact that isometric hip extension can occur below the pelvis constraints and possibly affect end-point kinetics recorded at the 6-DOF load sensor due to postural adjustments of the participant not being completely dissipated by the pelvis restraints. Since there is no quantitative evidence to support this, we deem it necessary to first observe the postural hip adjustments in a between group design before making the decision of whether we should control the level of hip extension occurring during the required protocol through the use of audio or visual cues. Thus, another benefit of this setup is having the ability to record such loads through our 1-DOF load cells placed under each leg. Regardless, we plan to corroborate recorded trunk kinetics with electromyographic recordings in all future work since this will permit us to quantify the level of coactivation that may occur that will not be registered by our kinetic measurements.

A final limitation is that errors can be induced through the technique of virtually translating the location of the 6-DOF load cell. Any measurement error and/or movement of the measured point during the experiment could affect the accuracy of the primary and secondary torques estimates being made with the larger concern being placed on the erroneous secondary torques. Therefore, at a minimum, a high-resolution 3-D digitization device should always be used to measure chosen 3-D coordinates of targeted spinous processes. Also, sensitivity analysis on measurement error should be performed on each cohort per study. This is necessary since the resulting errors are both influenced by the distance of translation and kinetics output of each subject. This was demonstrated between the relative differences of the single- and multi-axis isometric protocols and the subsequent decrease in error with voluntary secondary torques. Alternatively, errors arising from participant movement are harder to control but could be decreased through the use of continuous motion capture of bony landmarks or the postprocessing of strategically placed low-g accelerometers.

VI. Conclusion

In conclusion, the device and mechanical transformations discussed in this article will aid in the interpretation of isometric trunk kinetics from a wide range of populations. Our future goal is to use this device to investigate the effect of stroke on coupled trunk kinetics when compared to healthy controls.

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Biographies

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From 1988 to 2001 he worked as Pre-Doctoral Investigator, subsequently as a postdoctoral, clinical Assistant Professor and finally as a Senior Clinical Research Scientist in the Rehabilitation Institute of Chicago. From 2001 to 2005 he worked as tenure-track Assistant Professor in the Departments of Physical Therapy and Human Movement Sciences (PTHMS), Biomedical Engineering (BME) and Physical Medicine and Rehabilitation (PM&R) at Northwestern University. He became Chair and Tenured Associate Professor in PTHMS and Associate Professor in BME and PM&R, in 2006. In 2010, he became full Professor in PTHMS, BME, and PM&R. He is the Director of the neuroimaging and motor control laboratories and he is an Adjunct Professor and Chair of Neural Control and Rehabilitation, Department of Mechanical, Maritime and Materials Engineering, University of Technology, Delft, The Netherlands. His research interest encompass the characterization of neural mechanisms underlying motor impairments following brain injury due to stroke and cerebral palsy by using a combination of neuroscientific, engineering, and clinical sciences techniques.

Dr. Dewald is a member of the Society for Neuroscience, the American Physical Therapy Association, and the American Society of Biomechanics. He was recently appointed as a Fellow of the American Institute for Medical and Biological Engineering

Fig. 1.

Experimental device. Shown is the exploded view of the device (left), attached to a drawn participant (right).

Fig. 2.

Experimental device and orientation. Shown is a participant attached to the device (left) and the orientation of the 6-DOF load cell in relation to the device and anatomical planes of the attached participant (right).

Fig. 3.

Experimental apparatus attachment sequence. Shown is a demonstration of attaching a participant to the apparatus by first rotating the inner frame (A), securing the inner frame to the participant (B), restraining lateral movement of the pelvis and hip (C), as well as flexion/extension and anterior translation of the pelvis (D). Not shown is the lumbar support attachment.

$$\begin{bmatrix} 1 & 0 & 0 & 0 & 0 & 0 \\ 0 & 1 & 0 & 0 & 0 & 0 \\ 0 & 0 & 1 & 0 & 0 & 0 \\ 0 & dz & -dy & 1 & 0 & 0 \\ -dz & 0 & dx & 0 & 1 & 0 \\ dy & -dx & 0 & 0 & 0 & 1 \end{bmatrix} * \begin{bmatrix} fx \\ fy \\ fz \\ Mx \\ My \\ Mz \end{bmatrix} = \begin{bmatrix} fx \\ fy \\ fz \\ Mx_{sag} \\ My_{tran} \\ Mz_{cor} \end{bmatrix}$$

$$\begin{aligned} Mx_{sag} &= (dz * fy) - (dy * fz) + Mx \\ My_{tran} &= (dx * fz) - (dz * fx) + My \\ Mz_{cor} &= (dy * fx) - (dx * fy) + Mz \end{aligned}$$

Fig. 4.

Virtual translation. Example of a transformation matrix that can be used to virtually translate forces and moments from the 6-DOF load cell to the participant's assumed torque axis (left). The 3-D coordinates (dx, dy, dz) are in relation to the origin of the 6-DOF load cell.

Fig. 5. Single- and multi-axis isometric protocol results. Shown are the (A) torques (Mean $\pm$ SEM) generated in all planes during the single-axis protocol and the (B) normalized transverse torques generated during the multi-axis protocol (F-flexion, E-extension, RT-right twist, LT-left twist, 25–25% of max, 50–50% of max).

Fig. 6. Sensitivity to measurement error. Shown are the absolute error values (median $\pm$ range) as a % of maximum generated torque for the (A) single-axis protocol (flexion/extension—sagittal, left/right lateral flexion—coronal, left/right axial twist—transverse) and (B) multi-axis protocol (F-flexion, E-extension, RT-right twist, LT-left twist, 25–25% of max, 50–50% of max).

TABLE I
Single-Subject Force Data (Newtons) From Each 1-DOF Load Cell During Single-Axis Isometric Protocol

Direction	Flexion	Extension	Left	Right	Left Twist	Right Twist
Load Cell						
Lumbar	570.45	0.00	0.00	0.00	4.52	0.00
Left Leg	0.00	575.53	0.00	595.65	321.40	62.01
Right Leg	0.00	632.87	582.27	0.00	89.10	407.37

Due to participants not being rigidly attached to each of the 1-DOF load cells, only compressive loads can be measured. Therefore, posterior loads on the lumbar load cell and downward forces on each of the leg load cells are shown. If no compressive load was detected, the value was set to zero.