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Tracking Kinematic and Kinetic Measures of Sit to Stand using an Instrumented Spine Orthosis

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Abstract

Age related spinal deformity is an becoming increasingly prevalent problem, resulting in decreased quality of life. While spinal deformity can be corrected via surgical intervention, a large number of people with spinal fusions require follow-up surgery due to further degeneration. The identification of changes to a subject's kinematics and kinetics post-surgery are limited by a lack of methods to collect patient-specific motion data over the course of surgical recovery. This paper introduces an Instrumented Spine Orthosis (ISO) that can capture the motions of the subject torso without requiring the use of a control computer or other dedicated motion capture equipment. This system is used to collect the peak torso angles and velocities for a single subject performing sit-to-stand actions. The accuracy of the ISO is evaluated using motion capture, during different sit-to-stand protocols designed to highlight motion changes that have been seen in subjects with reduced mobility. This system was found to provide reliable measurements of these kinematic and kinetic torso measures across all tested motions, demonstrating the potential for the use of Instrumented Spine Orthotics to provide quantitative measures during the surgical recovery process.

I. INTRODUCTION

Adult spinal deformity and scoliosis is found in approximately 68% of the population over the age of 60 [1]. In subjects with osteoporosis or osteopenia, there is a risk of accelerated degeneration, resulting in pain and the reduced load capacity of the spine. Individuals with spinal deformity can be supported through the use of orthotic braces, conditioning exercises, and surgical procedures such as decompression and stabilisation via spinal fusion [2]. Unfortunately, roughly 40% of subjects who undergo spinal fusion can have subsequent Proximal Joint Kyphosis and Fracture (PJK, PJF) above the fused section within 18 months of their operation, requiring surgical revision [3].

Investigations into the underlying causes of PJK and PJF are limited to static assessments based on radiological techniques [4]. While these methods can provide estimates of vertebral loading for a single posture, they lack the ability to model the effect of fusion on the peak vertebral loads during movement. A quantitative method to assess and track motion pre- and post-surgery would allow for increased understanding of the effect of spinal fusion on the dynamic loading of the vertebrae.

A. Quantitative Measures of the Spine

Quantitative measurements of the spine are typically obtained through optical/magnetic tracking methods [5]. These methods can provide 3D reconstructions of multiple regions of the spine, allowing for detailed analysis of motion. Through the support of radiographic techniques, predictive models of spinal motion have been developed for computational musculoskeletal analysis [6], [7].

Inertial Measurement Units (IMUs) have been used in the biomechanics community for estimation of spine motions. Tafazzol [8] used three commercial IMU sensors (Xsens) mounted on T5, L1, and S1 to estimate lumbo-pelvic rhythm. The same sensors (located at C1, T1, L4) have also been used to estimate the rotations of the head, torso, and hip [9] and the 3D relative rotations in the upper and lower thorax, lumbar spine, and pelvis [10]. These works have been validated through the use of optical motion capture, or direct goniometry, and have been shown to provide precise measurements of spine motion. While these IMU systems are more portable than conventional motion capture methods the expertise and cost to use these methods still limits their clinical deployment outside of specialist biomechanical facilities.

An alternative approach to the measurement of clinical metrics is the development of wearable devices in conjunction with functional movement tests. Marras [11] was able to distinguish between 374 healthy, and 335 subjects with Low Back Disorders using a triaxial spine exoskeleton and an extensive set of range of motion exercises. The sagittal range of motion, and peak flexion/extension velocities and accelerations were found to be indicators of structural or muscular based low back pain. This *functional performance score* has been used in longitudinal tracking of subjects with low back pain [12] and the prediction of low back pain outcome measures [13].

B. Contributions

This work demonstrates an accurate, portable, and affordable system to measure key kinematic and kinetic variables by instrumenting a commercial back brace with IMUs. These spine braces are frequently given to spinal fusion patients post-surgery allowing for unobtrusive collection of motion data in clinical cohorts. The use of these instrumented braces allows for the longitudinal tracking of spine function and surgical recovery outside of dedicated biomechanics labs and motion capture facilities.

Instead of attempting to perform a detailed kinematic reconstruction of the spine, this work aims to extract a number of key biomechanical markers that highlight compensatory strategies being used by the subject. The utility of this technique is demonstrated through the analysis of sit-to-stand.

C. Biomechanics of Sit-to-stand

The Sit-to-Stand (STS) action is a well studied biomechanical action due to its importance in daily living and as a functional test of motor control and stability. A successful STS action requires significant muscular strength and coordination, allowing for the identification of limitations due to impairment.

The motions of a subject performing STS can be separated into four distinct phases [14], [15]:

1. *Momentum generation* is the initial flexion movement of the trunk and pelvis generating upper-body momentum.
2. *Momentum transfer* occurs when the upper-body momentum generated in phase 1 is transferred to the whole body, resulting in an anterior and upward acceleration.
3. *Deceleration* is the rising motion of the subject which can be observed as the extension at the hip and knee joints from the crouched posture of phase 2 to the standing posture of phase 4. This is coupled with the deceleration of the torso.
4. *Stabilisation* is the final state of the standing action, where the subject actively balances to remain in the standing state.

While healthy subjects appear to follow these phases, alternative strategies have been identified. Subjects can instead stand *quasi-statically* (referred to as stabilisation by Hughes [16]). In quasi-static standing, the subject shifts their centre-of-mass over their base-of-support (feet). This action requires less momentum for a successful stand, instead relying on repositioning of the torso and/or feet.

The strategy chosen by a subject has been linked to the subject's ability to extend their knee [16]. In momentum transfer, the momentum of the torso assists the knee musculature in performing the standing action. The developed momentum must be sufficient to allow for transfer to the standing state, without overshoot [17]. The quasi-static strategy does not require this precise control, reducing the unstable momentum transfer phase through repositioning. These two strategies can be combined by subjects, reducing the required momentum demand though repositioning the body over the feet, before performing a separate rise. This has been seen in able-bodied elderly subjects who exhibit increased trunk flexion and a higher trunk velocity when compared to younger subjects [18].

II. METHODOLOGY

This section outlines the development of the instrumented spine orthosis and the mathematical modelling used to recover the kinematic and kinetic states of the spine.

A. Instrumented Spinal Orthosis (ISO)

A Spinomed® IV A/P spinal brace was used as the base spinal orthotic in this study [19]. This orthotic is indicated for use for hyperkyphotic subjects and subjects with osteoporitic bone collapse and was selected based on its use in prior studies and from consultation with Orthotic and Prosthetic specialists. This spinal orthotic consists of a cold-workable splint that is shaped to follow the desired curvature of the spine and a set of belts to secure the splint onto the subject.

Two IMUs (LSM9DS1, Sparkfun) were attached onto the splint using Velcro® coins at the C7/T1 and L5/S1 levels. These IMUs were connected to an ATSAMD21G18 ARM Cortex M0 processor (M0 Feather, Adafruit) via an I2C multiplexer. Raw magnetometer, gyroscope,

and accelerometer data were timestamped using a PCF8523 Real-Time Clock and stored on an integrated 8GB microSD card via a serial connection. The electronic package is powered by a 6.6Ah lithium ion battery, with a CR1220 coin cell to continually power the RTC. Based on an expected filesize of 200kB per dataset, the device is power-limited, with an expected operational time of approximately 3 months between charges.

B. Mathematical Framework

The kinematic and kinetic recovery of the torso can be expressed as a geometric rigid-body state estimation problem [20]. The splint of the back brace allows the IMUs at C7/T1 and L5/S1 to act as two sets of observations for the same rigid-body. As there are no direct observations of the lower limbs, the torso is modelled as a floating link that can rotate axially and in the flexion/extension direction. While the focus of this work is the recovery of the flexion/extension of the spine, additional degrees of freedom are added to account for variations in the initial orientation of the subject, and to account for the motions of the prior body segments (Figure 1). This model can be expressed mathematically as:

$$\mathbf{g}_{WT}(\theta_Z, y, z, \theta_R, \theta_{FE}) = \mathbf{g}_{WP}(\theta_Z, y, z) \mathbf{g}_{PT}(\theta_R, \theta_{FE}) \quad (1)$$

where $\mathbf{g}$ denotes a rigidbody rotation between two frames, and W , P , and T represent the world, pelvic, and thoracic frame respectively.

Each of these states can be related to their first three derivatives in discrete time using the forward Euler method. For each of the states x_k and the full kinetic state $\mathbf{X}_k$, the state at the next sample $k + 1$ can be expressed as:

$$\begin{aligned} \mathbf{X}_{k+1} &= \begin{bmatrix} x_k + 1 \\ \dot{x}_k + 1 \\ \ddot{x}_k + 1 \end{bmatrix} \\ &= \begin{bmatrix} 1 & \delta t & (\delta t)^2/2 \\ 0 & 1 & \delta t \\ 0 & 0 & 1 \end{bmatrix} \mathbf{X}_k + \begin{bmatrix} (\delta t)^3/6 \\ (\delta t)^2/2 \\ \delta t \end{bmatrix} \ddot{x}_k \end{aligned} \quad (2)$$

where δt is the time-step between samples. As the effective joint torques are unknown, Equation 2 can be represented as a random noise process driven in the jerk term $\ddot{x}$.

These states and their derivatives can provide an expression for the torso body velocities $\mathbf{V}_{W,T}^b$ and accelerations $\dot{\mathbf{V}}_{W,T}^b$ via a set of forward recursive equations [21].

The IMUs are modelled as having a fixed location and orientation offset from the torso. The homogeneous transform from the torso frame to each IMU frame $\mathbf{g}_{T,IMU}$ can be expressed by the twist parameters ξ . These parameters can be identified through a calibration step or

through direct measurement. After these parameters have been identified, the IMU body velocities and accelerations can be written explicitly using the Adjoint operator Ad [21]:

$$\mathbf{V}_{IMU}^b = \begin{bmatrix} \mathbf{v}_{W,IMU}^b \\ \boldsymbol{\omega}_{W,IMU}^b \end{bmatrix} = Ad_{\mathbf{g}_{T,IMU}^{-1}} \mathbf{V}_{W,T}^b \quad (3)$$

$$\dot{\mathbf{V}}_{IMU}^b = \begin{bmatrix} \dot{\mathbf{v}}_{W,IMU}^b \\ \dot{\boldsymbol{\omega}}_{W,IMU}^b \end{bmatrix} = Ad_{\mathbf{g}_{T,IMU}^{-1}} \dot{\mathbf{V}}_{W,T}^b \quad (4)$$

The raw IMU readings of compass heading, angular velocity, and linear acceleration are modelled as the noisy measurements $\mathbf{y}_k$ of a normalised compass direction $\boldsymbol{\Omega}_W$, and the body velocities and accelerations:

$$\mathbf{y}_k = \begin{bmatrix} \boldsymbol{\Omega}_{IMU} \\ \boldsymbol{\omega}_{IMU} \\ \dot{\mathbf{v}}_{IMU} \end{bmatrix}_k = \begin{bmatrix} \mathbf{g}_{W,IMU}^{-1} \boldsymbol{\Omega}_W \\ \boldsymbol{\omega}_{W,IMU}^b \\ \dot{\mathbf{v}}_{W,IMU}^b \end{bmatrix}_k + \mathbf{n}_k \quad (5)$$

where $\mathbf{n}$ is taken to be normally distributed noise.

Given the process model (Equation 2) and observation model (Equation 5), the two IMU sensors can be fused using an Unscented Kalman Filter [22] to provide an estimate of the system state.

III. EXPERIMENTAL VALIDATION

This section outlines the experiments used to demonstrate the feasibility of tracking torso motions using the instrumented spine orthotic.

A. Standing Strategies

Four different standing strategies were investigated, each corresponding to the strategies and methodology listed in Section I-C.

1. *Self-selected* standing speed was the first set of experiments where the subject stood without direct coaching.
2. *Quasi-static* attempted to reduce the contribution of momentum transfer on standing. The subject was asked to stand at a speed where they could stop mid motion at any time without falling.
3. *Momentum Transfer* attempted to increase the contribution of momentum transfer on the standing action. The subject was asked to stand as fast as possible.

4. *Full-flexion* emphasised the flexion phase of the motion. The subject was asked to stand after fully flexing their torso while seated before standing, shifting their centre of mass as far forward as possible.

B. Experimental Procedure

A single asymptomatic subject was tested in this study (M, aged 29, 1.78m, 65kg) under informed consent (UC Berkeley IRB:2015-07-7767). The ISO was fitted to the subject by bending the splint section to the subject's standing lordosis. Active motion capture markers (Phasespace) were placed on the subject using a modified Plug-in-Gait protocol [23]. Additional markers were placed at S1 and L4/L5 to ensure at least 3 pelvic frame markers were visible in each frame. The ASIS and PSIS markers were located on the waist-band of the brace instead of directly on the skin (Figure 2).

The subject performed three sets of three sit-to-stand actions for each standing strategy after being trained to perform each action. The subject crossed their arms with their fingers touching their shoulders, and elbows on their chest, to reduce the effects of arm swing on the standing action. Active motion capture data was sampled at 480Hz, with the ISO logging IMU measurements at approximately 24Hz.

These measurements were processed offline using MAT-LAB. The CLAV, STRN, C7, T10, and SHO markers were used to define the rigid-body torso frame, while the ASIS, PSIS, S1, and L4/L5 markers defined the pelvic frame. These frames were defined using the principal axis definitions suggested by the ISB [24], [25]. The origins of these frames were shifted as seen in Figure 1 to reduce the effect of single marker error on the rigid-body recovery process.

A ground truth estimate of model states was obtained from these rigid-body frames using the analytic Padan Kahan solutions for inverse kinematics [26]. These model states were compared to the state estimates recovered by the ISO.

C. Results

The recovered joint angles from the motion capture data were repeatedly filtered (lowpass Butterworth 3Hz) and numerically differentiated to obtain estimates of the angular velocities and accelerations. The motion capture and IMU data were time aligned, and the filtered motion capture state sub-sampled at the same times as the IMU data. The time trace for normal sit-stand-sit cycle is shown in Figure 3.

The peak values for the sit-to-stand segments re tabulated for each STS strategy (Table I). The peak-to-peak absolute error between the IMU and motion capture states are given.

D. Discussion

The proposed method for tracking torso flexion/extension was found to provide comparable estimates of peak flexion/extension angles and velocities. The peak angle and angular velocity estimates were found to be reliable for the self-selected, quasi-static strategies, with MAE errors of under XXX degrees and XXX degrees per second.

The MAE of the flexion/extension angle increased in the full-flexion strategy. This increase may be due to the large amount of soft tissue deformation at this high flexion angle.

In the momentum strategy, the MAE of all three metrics increased. This decline in performance may be due to the inaccuracies in the observation model becoming more noticeable at the higher angular velocities and accelerations.

The utility of tracking the proposed kinematic and kinetic measures can be seen in the change between self-selected standing and the three alternative strategies. In quasi-static standing, a subject reduces the momentum transferred in phase 2 standing by reducing the initial flexion angle and peak flexion velocity in phase 1. This is seen as a statistically significant decrease in the angles and angular velocities seen in the motion capture and IMU data (motion capture $p = 1 \times 10^{-8}$, IMU $p = 1 \times 10^{-9}$ respectively).

This can be contrasted by the momentum standing action, where a subject attempts to increase the momentum generated and transferred in phases 1 and 2. The peak flexion angular velocities were found to be significantly higher than in self-selected standing ($p = 1 \times 10^{-11}$ for both sensing modalities).

In full flexion standing where the subject was asked to fully flex their torso before standing-moving their centre of mass as far forward as possible before standing. While the peak flexion angle was found to increase significantly (motion capture $p = 1 \times 10^{-11}$, IMU $p = 1 \times 10^{-7}$ respectively), the angular velocities were not found to change.

IV. CONCLUSION

This paper introduces and demonstrates the potential for using an instrumented spinal orthotic device to quantify and track kinematic and kinetic motion biomarkers. While the feasibility of using this device has been shown, there are a number of limitations to this study. Additional healthy subjects need to be tested to determine the inter-subject variability of these measurement, and standing strategies. Clinical subjects need to be tested with the ISO to determine if the hypothesised compensation strategies are representative of this population, and if these methods could be used for longitudinal tracking. The biomechanical model of the subject wearing the back-brace is highly simplified with the torso and abdomen assumed to move as a single link due to the orthotic. The validity of this assumption needs to be tested especially if softer braces are used.

A. Future Work

The developed ISO system is scheduled for testing with an age- and sex-matched cohort of spinal fusion subjects and controls. The metrics identified in this paper will be tested before and after a short walk period. This walk test is included based on recent findings that a short walking period of 10 minutes was sufficient to reduce any compensatory mechanisms in the spine due to fatigue[27]. This procedure may aid in the longitudinal identification and tracking of patient recovery.

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Fig. 1.

Wireframe of a standing subject showing the coordinate frames and states used in this model. The transform between the world and the pelvic frames can be written as an initial rotation about the Z axis (θ_Z) and a translation in the Y,Z plane (y, z). The transform from the pelvic to the thoracic frame can be written as a rotation about the X axis (flex-extend θ_{FE}), a rotation about the Z axis (torso rotation θ_R) and a fixed position offset.

Fig. 2.

Rear and right views cartoons of subject wearing the ISO. ISO spinal splint is shown in white, with the strapping belts in light grey. Active motion capture markers are shown as red crosses. IMUs (blue circles) are attached to the ISO at the locations shown.

Fig. 3.

A representative trace of torso flexion angle and angular velocity for self-selected sit-to-stand. Joint states recovered using active motion capture and the ISO are shown in dotted and solid lines respectively. All three sit-stand-sit reps are shown, with the stand-to-sit portions greyed out.

Key kinematic and kinetic variables recovered during sit-to-stand for the four different standing strategies. The mean and standard deviations of each measure are given for each sensing modality. Cells in bold are statistically different from self-selected standing ($p < 0.01$).

TABLE I

	Peak Flexion Angle (deg)			Peak Flexion Velocity (deg/s)			Peak Extension Velocity (deg/s)		
	Mocap	IMU	Abs. Error	Mocap	IMU	Abs. Error	Mocap	IMU	Abs. Error
Self-selected	54.9±4.4	53.5±5.9	2.1±1.9	103±6	94±5	9±3	86±17	87±14	3±3
Quasi-static	22.3±2.8	23.2±4.0	2.0±1.6	47±13	42±5	5±9	22±4	20±4	2±2
Momentum	66.3±3.4	62.4±5.5	5.9±2.7	182±15	157±10	25±6	165±33	141±21	24±14
Full Flexion	86.8±4.1	83.5±8.1	5.9±3.6	84±23	74±19	10±6	88±14	85±12	4±2