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Quantification of the development of trunk control in healthy infants using inertial measurement units

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Abstract— **Trunk motor control is essential for the proper functioning of the upper extremities and is an important predictor of gait capacity in children with delayed development. Early diagnosis and intervention could increase the trunk motor capabilities in later life, but current tools used to assess the level of trunk motor control are largely subjective and many lack the sensitivity to accurately monitor development and the effects of therapy. Inertial measurement units could yield an objective quantitative assessment that is inexpensive and easy-to-implement. We hypothesized that root mean square of jerk, a proxy for movement smoothness, could be used to distinguish age and thereby presumed motor development. We attached a sensor to the trunks of six young children with no known developmental deficits. Root mean square of jerk decreases with age, up to 24 months, and is correlated to a more established method, i.e., center-of-pressure velocity, as well as other standard inertial measurement unit outputs. This metric therefore shows potential as a method to differentiate trunk motor control levels.**

I. INTRODUCTION

A. Motivation

Approximately half of typically developing children in the Netherlands can sit unsustained without arm support at 7.5 months and are able to walk without support at 15 months [1]. These motor milestones are important indicators for later motor skill acquisition [2]. Trunk motor control is a predictor of gait capacity in populations with delayed development, such as children with cerebral palsy [3].

Neuromotor disorders such as cerebral palsy can not only cause delays in motor milestones [4], but can also impair the child's gross motor function in the long term, essentially limiting the child's ability by the age of five [5]. Early intervention, e.g., in the first four years, has the potential to increase the trunk motor control abilities of affected children later in life [2]. While there has been rapid progress in evidence-based intervention in clinical practice [6], there is still a need for sensitive assessment tools that can accurately identify children who would most benefit from these therapies and that can effectively track progress.

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Current clinical tools to assess trunk motor control in seated children are subjective and observation-based. Nine tools have been tested to assess sitting balance, are used in children with neuromotor disorders, and have sufficient information on clinimetric properties [7]–[14]. Of these, only a subset have been tested for young children below the age of five [7]–[12].

Most of these tools evaluate a combination of static and active balance. Static balance reflects the ability to remain upright without movement; active balance reflects the ability to stay balanced while inclining or rotating the trunk or while moving the extremities [15], [16]. However, none of these tools have established large-scale implementation in clinical practice.

The aim of this study was to analyze an objective metric, i.e., the jerk of trunk movement, to quantify different levels of trunk motor control. Trunk movement was measured with an inertial measurement unit (IMU). Quantifying movement with an IMU, rather than traditional biomechanical sensors such as optical motion capture or force plates, is inexpensive and easy to implement, facilitating clinical uptake. Such an assessment may minimize subjectivity between evaluations and could potentially distinguish small changes in trunk motor control [17].

B. Related work

For objective assessment, sitting balance is most commonly quantified by measuring center of pressure (COP) with a force plate [18]–[20]. COP metrics have shown good reliability in healthy children under the age of one [21] and in children under the age of two with or at risk of cerebral palsy [18]. Force plates have high accuracy, but are generally expensive and restrict the measurements to a single location [22], issues that other types of sensors, such as magnetic trackers [23], [24] and optical systems [17] used in assessing trunk motor control in young children in the sitting position also suffer from.

IMUs are relatively cheap, easy to use, and portable. These sensors have been used to measure seated balance [25], and, more commonly, for standing balance though only in adult populations [26].

Jerk of trunk movement has been used to quantify postural balance in quiet stance in both Parkinson's disease patients [27] and Huntington's disease patients [28], where it was able to distinguish between healthy and affected participants. Furthermore, a recent study showed that the jerk in quiet stance decreased with age for participants from five years old to adulthood [26]. As a metric for smoothness of sway [27], we hypothesize that a jerk-based measure could be used to track motor development of trunk control during quiet sitting.

C. Research questions

This preliminary research aims to investigate the following primary research question: **Can a trunk-attached IMU be used to differentiate between different levels of trunk motor control in typically developing young children with different ages, e.g., of up to four years?**

In order to answer the primary research question, the following secondary questions will be explored:

1) Is the RMS of jerk (RMS(j)), as determined from a trunk-attached IMU, an effective outcome metric for differentiation between different levels of trunk motor control?
2) Does this IMU-based metric generate similar results as a more established method, e.g., COP velocity?
3) Are there other candidate metrics that can be derived from the outputs of a trunk-attached IMU?

II. Methods

A. Participants

Eight children were recruited, such that each child was between six months and four years old and had no known health problem. The parents of the children signed an informed consent form. The informed consent form and the experimental procedures were both approved by the Human Research Ethics Committee of the TU Delft.

Because we only considered quiet sitting in this study, children who could not sit still for at least 30 seconds were excluded from further analysis. One child became restless and agitated when we tried to apply the instrumentation to him. We terminated the experiment with this child and thus excluded this child from the rest of the study. Another child could not sit still for the duration of the experiment, running around the collection space. The quiet sitting that did occur for this child was not sufficiently long for our chosen metric, so this child was excluded as well.

Each child participated in a single experimental session with one of their parents. The session consisted of several tasks, including quietly sitting on the ground and on a bench and goal-directed movements, of which only quiet sitting was analyzed for this study.

B. Experimental setup

Fig. 1 presents an overview of the experimental setup, which consisted of

- An Xsens MTw Awinda IMU: a tri-axial sensor that combined data from an accelerometer, a gyroscope, and a magnetometer [29] to obtain an orientation estimation of the sensor.
- Kistler type 9260AA6 force plate: platform that measured the ground reaction forces and determined the point of application, i.e., the COP.

The IMU was attached to a black elastic band that was wrapped around the trunk, just below the armpits, so that the IMU was placed near where the center of mass was estimated. The child was then placed in a seated position on the force plate. The IMU and force plate data are synced via an external trigger, sampled at 100 and $500\,\mathrm{Hz}$, respectively.

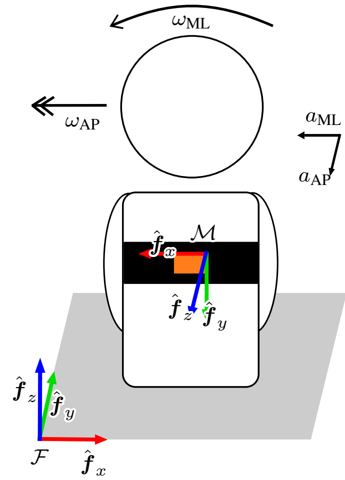

Fig. 1: Schematic overview of the experimental setup. The IMU was attached by an elastic band wrapped around the trunk below the armpits. The child quietly sat on a force plate, depicted in gray. Jerk (j) and acceleration (a) were assessed in the anteroposterior and mediolateral directions, or the $\hat{\boldsymbol{m}}_z$- and $\hat{\boldsymbol{m}}_x$-directions, respectively. Anteroposterior (ω_{AP}) and mediolateral (ω_{ML}) angular velocities were assessed as shown in the figure as well.

The coordinate frames, shown in Fig. 1, were separately defined for the IMU and the force plate. The $\mathcal{M}$-frame was a body-fixed frame with the origin placed in the top-right corner of the sensor, corresponding with the standard sensor coordinate system [30]. The $\mathcal{F}$-frame was positioned on the force plate with the origin on the back left corner. The vertical axes were aligned, but rotation was possible in the transverse plane, e.g., if the child turned during the trial. However, this did not impact analysis, as only the magnitudes of the COP velocity and the IMU-based measurements were compared.

C. Data pre-processing

An overview of the data processing can be found on the left side of Fig. 2. Gravity was removed from the raw accelerometer data, by first computing gravity in the $\mathcal{M}$-frame, using the rotation matrices provided by the onboard algorithm [30] and then subtracting the gravity component from the accelerations measured by the device. The adjusted accelerations and raw gyroscopic data were filtered using a second-order low-pass Butterworth filter, with a cut-off frequency of $10\,\mathrm{Hz}$. A pilot experiment determined that $90\,\%$ of the total power of the movements of another infant lay below $10\,\mathrm{Hz}$, and this threshold was confirmed for the children measured in this study. The same filter was used for the force plate data.

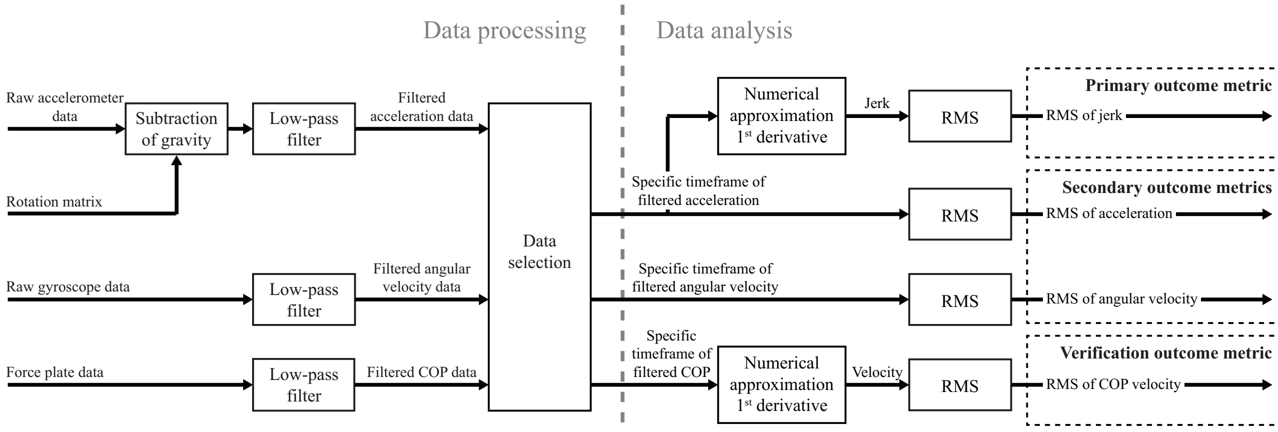

Fig. 2: Graphical overview of the data processing and analysis structure.

D. Data selection

After compensating for gravity and filtering the data for the entire trial, the time frame used in our analyses was selected. Prior to data selection, data were excluded where

- The child is not on the force plate, or
- The parent holds the child.

Data from an optical motion capture system were inspected to manually identify these events. After exclusion of unusable data, the 30-second segment with the minimum RMS of the resultant of the angular velocity was chosen for further analysis. The segments chosen by this method most often corresponded to manually identified periods of quiet sitting, and the segment length is standard practice in other clinical metrics [31]. We were also interested in the reproducibility of this data selection algorithm, so three non-overlapping, 30-second segments were selected for an additional analysis, presented in the supplemental results.

E. Data analysis

The right side of Fig. 2 shows an overview of the data analysis structure.

1) Primary outcome metric: RMS of jerk (RMS(j)): Jerk, or the first derivative of acceleration, is an indication of movement smoothness [27]. The RMS of jerk has shown high discriminative ability in the assessment of postural control in Parkinson's disease patients [32].

Jerk was computed as a function of the processed accelerations measured by the IMU. The derivative of acceleration was approximated using the following fourth-order central-difference approximation,

$$f'(x) = \frac{8f(x+h) - f(x+2h) + f(x-2h) - 8f(x-h)}{12h}, \tag{1}$$

where f represents a general function of x and h represents the sampling time. The RMS was then computed in each direction and for the resultant vector in the transverse plane, denoted RMS(j). Because we were interested in trunk control as a metric for sitting balance, we excluded the vertical component as we expect that height modulation during quiet sitting is less relevant to maintaining an upright posture. A paired t-test was computed to analyze differences in the RMS of jerk in the two directions.

2) Verification outcome metric: RMS of COP velocity: The RMS of COP velocity, denoted RMS(v_{COP}), was used to verify the RMS of jerk and the secondary outcome metrics described below. COP velocity is commonly used as a metric for postural control in quiet stance [33]. The RMS of COP velocity is expected to decrease with an increase in trunk motor control [33]. The COP velocity was computed by Eq. (1) to numerically approximate the derivative of the COP position, measured by the force plate. All comparisons were made to the resultant of the COP velocity, i.e., the magnitude of the COP velocity in the plane of the force plate ($\hat{\boldsymbol{f}}_x$-$\hat{\boldsymbol{f}}_y$). Pearson correlation coefficients were computed in Matlab, using the standard statistical toolbox.

3) Secondary outcome metrics: The included secondary outcomes are the RMS of acceleration and the RMS of angular velocity. Linear accelerations and angular velocities are standard outputs of the IMU system, so they are not subject to the numerical inaccuracies of differentiation. We analyzed the RMS of acceleration in the anteroposterior (AP) and mediolateral (ML) directions, or the $\hat{\boldsymbol{m}}_z$- and $\hat{\boldsymbol{m}}_x$-axes, and the RMS of angular velocity for AP and ML rotations, as shown in Fig. 1. The RMS of linear acceleration was denoted RMS(a_{AP}) and RMS(a_{ML}) for the AP and ML directions, and the RMS of angular velocity was denoted RMS(ω_{AP}) and RMS(ω_{ML}) for the AP and ML directions, respectively.

F. Data availability

Data and code to replicate the results, as well as the supplemental results, can be found at doi.org/10.4121/19236381. All calculations were done in Matlab (R2019b).

III. Results

The RMS of jerk indicated a downward trend for children aged 10–24 months in both the resultant and in the AP and ML directions (Fig. 3). The RMS of jerk for the children aged 13 months ($\circ$, $\times$) were comparable. The RMS of jerk in the ML direction was higher for all children in this study than the RMS of jerk in the AP direction (paired t-test, $t(5) = -2.69$, $p = 0.04$).

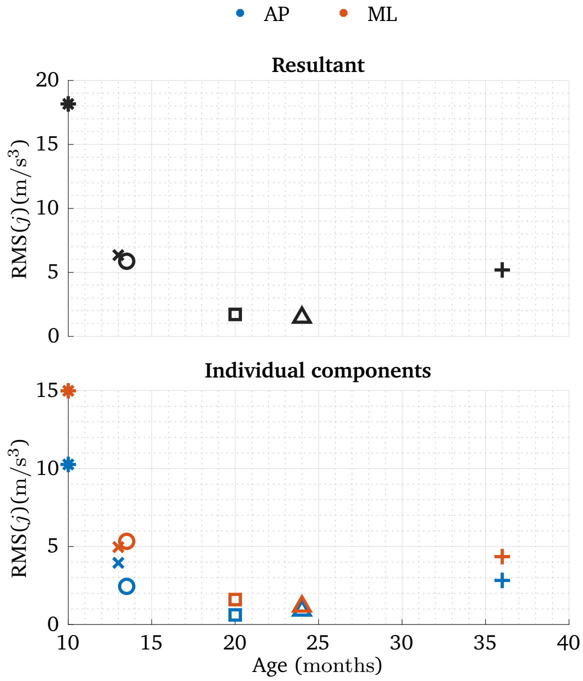

Fig. 3: RMS of jerk for 30 seconds of quiet sitting. There was a downward trend for children up to two years of age, with an increase for the oldest child. In general, the RMS of jerk in the ML direction was higher than in the AP direction.

In general, the RMS of jerk did not vary substantially across three different segments of quiet sitting (Supplemental Fig. S1). The lowest RMS of jerk often corresponded with the minimal RMS of angular velocity, with some exceptions that did not affect the overall downward trend.

The RMS of COP velocity exhibited a similar downward trend for these children up to 24 months of age (Fig. 4). A Pearson correlation coefficient was computed between the RMS of the resultant jerk and the RMS of the resultant COP velocity, indicating a positive correlation ($r(4) = 0.88$, $p = 0.02$). Similar computations were done for the RMS of jerk in the two directions, similarly indicating a positive correlation for the AP direction ($r(4) = 0.90$, $p = 0.01$) and for the ML direction ($r(4) = 0.86$, $p = 0.03$).

The RMS of linear acceleration and the RMS of angular velocity also exhibited a downward trend for the first two years of age (Fig. 5). The RMS of linear acceleration in both the AP and ML directions were positively correlated with the RMS of COP velocity ($r(4) = 0.99, 0.89$, $p < 0.01, p = 0.02$, respectively), as were the RMS of angular velocity in both the AP and ML directions ($r(4) = 0.98, 0.82$, $p < 0.01, p = 0.04$, respectively).

IV. Discussion

A. RMS of jerk during quiet sitting as a metric for trunk motor level differentiation

The RMS of jerk is expected to decrease with increased motor control, as movement smoothness is expected to increase [27]. This downward trend is visible in Fig. 3 for ages 10–24 months. Furthermore, following the hypothesis that children with approximately the same age have the same developmental level, the similarity in magnitude for the 13-month-old children ($\circ, \times$) is encouraging. These results indicate that this metric could be a viable candidate as a method to track developmental age.

The increased RMS of jerk for the oldest child ($+$) was an unexpected outcome. This child had a high motor capacity, but fidgeted during the measurements. Jerk-based measures are sensitive to starting and stopping periods [34]. The metric was also not designed to distinguish intentional movements, so this work could benefit from either a method to ensure quiet sitting, especially in children with high motor capacity, or a metric that is agnostic to the movement condition.

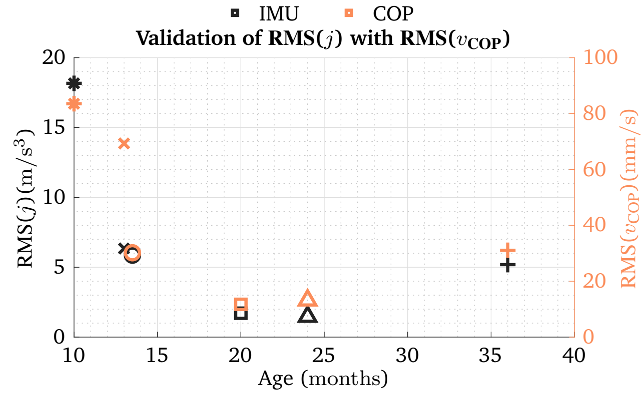

Fig. 4: RMS of jerk compared to RMS of COP velocity. The RMS of COP velocity followed a similar trend to the RMS of jerk, and the two were positively correlated.

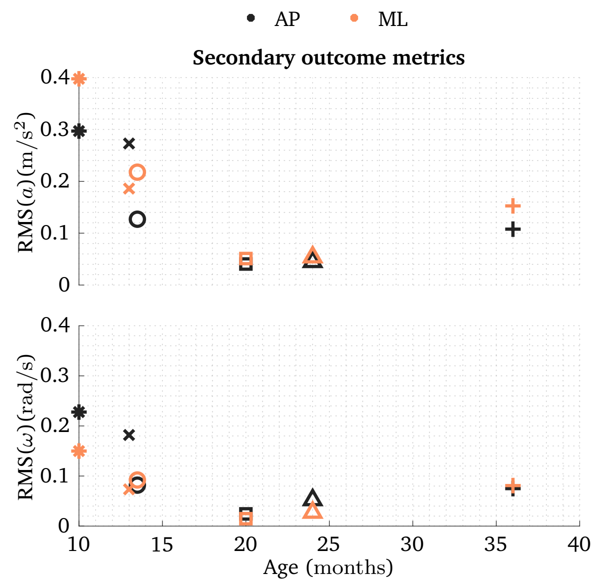

Fig. 5: RMS of linear acceleration and RMS of angular velocity in both the AP and ML directions. These metrics exhibited similar downward trends for the first two years.

Across participants, the RMS of jerk in the ML direction was greater than in the AP direction. If a higher RMS of jerk correlates with poor trunk control as our results suggest, then this could indicate that the AP and ML directions develop at different rates, with ML control developing more slowly than AP control. Although differences have been found with respect to AP and ML postural control development [20], [23], more research is needed to understand how motor control interacts in these directions.

The RMS of jerk also appeared to be agnostic to the chosen data segment. Measuring movement in this population is difficult, as the children are less likely to follow explicit instructions or to stay focused for the duration of the trial. For an exploratory study, the similarity in the trend for the different data segments is promising. In future work, we will perform sensitivity analyses to determine the reliability of this metric and define clear recommendations for how to collect an accurate measurement for this population.

B. Comparison with the RMS of COP velocity

The RMS of COP velocity shows a similar trend to the RMS of jerk, indicating that the chosen metric may represent a similar phenomenon to the more established technique of tracking the RMS of COP velocity. The RMS of jerk was positively correlated with the RMS of COP velocity in both the AP and ML directions and for the resultant vector. Differences in the correlations across the two directions are likely to be the result of the discrepancy between the RMS of the COP velocity for the second youngest child ($\times$). Given the small sample size, we cannot draw strong conclusions on how this difference arose. A post-hoc review of the data indicated that that child incrementally slid forward in a rocking motion. This may be an advantage for using the RMS of jerk rather than of COP velocity, in that the jerk is insensitive to these low-frequency movements. More rigorous experiments should be conducted to determine the exact nature of the relationship between the RMS of jerk and the RMS of COP velocity.

C. Other outcome metrics to track motor development

The RMS of linear acceleration and the RMS of angular velocity in this sample exhibit similar trends to what was observed for the RMS of jerk. While jerk has clinical relevance as a proxy for smoothness, acceleration and angular velocity require less post-processing as they are direct outputs of the IMU and do not need to be differentiated, like jerk or the COP velocity. The discrepancy in the angular velocity between the two directions in the second youngest child ($\times$), especially compared to the relationship between the AP and ML directions for the other children, could indicate that there are other factors that contribute to the angular velocity that may make this a less suitable metric for evaluating trunk control. A trunk control metric should be a measure of the child's development at that point in time, rather than a value that can change depending on the movement in question, and we expect acceleration and velocity to more closely align with specific movements than the overall quality of movement [35]. More children should be examined while undergoing different movements to determine if these differences are significant or an artifact of our experimental protocol. For future work, a larger range of outcome metrics could be systematically compared for a larger, longitudinal sample.

D. Study limitations

As an exploratory study, this work is inherently limited by the sample size and sample characteristics. With only six children across a broad age range, we cannot confidently generalize these results to all healthy children. Furthermore, each child was only measured for one session, so we cannot make any claims for how these values change as the child develops. Additionally, the age range chosen for this study was based on the gap in existing metrics, but was perhaps too broad for this metric. Most typically-developing children can walk by 18 months [36], which necessarily means they have adequate trunk motor control. We therefore plan to replicate this study longitudinally for a larger number of children across a smaller age range (between 6–12 months). This new study could describe changes in trunk motor control related to an individual's development and could increase the generalizability for the target age range.

In this analysis, we used the chronological age rather than the developmental age. Before the experiment, we asked parents to complete the van Wiechen scheme [37], a common method to determine developmental age in the Netherlands. Because parents were not trained and we were unable to have therapists complete the scheme, we did not use these values for analysis. While we found that the determined developmental age was close to or slightly above the chronological age for all children, using the developmental age rather than the chronological age will be important in assessing children with neuromotor disorders, for whom the two ages could widely differ.

Finally, we want to replicate these experiments with patient populations. While it is important to establish the baseline in healthy children in this age group, we are interested in how these curves change with neuromotor disorder, such as cerebral palsy, and if these metrics can be improved with rehabilitation.

V. Conclusion

This study is the first to use outcome metrics computed from a trunk-attached IMU as an objective tool for the assessment of trunk motor control in children under four years old. While further study is still required, the preliminary results for the RMS of jerk as an outcome metric are encouraging for ages of 10–24 months, where a downward trend was visible. The metric appeared to be relatively independent of chosen data segment and compared well to the RMS of COP velocity, a more established method.

However, with a sample size of only six children, these results are preliminary. A follow-up longitudinal study is advised to further examine the effectiveness of the RMS of jerk as an outcome metric for the differentiation of trunk motor control levels.

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