

## Original Article

# Development of a multibody model to assess efforts along the spine for the rehabilitation of adolescents with idiopathic scoliosis

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**Introduction:** Gait analysis has often been recognized as helpful for the therapeutic follow-up of adolescents with idiopathic scoliosis (IS). **Methods:** A multibody model of the human body was developed to display the intervertebral efforts along the spine of each adolescent with IS, and highlight the efforts that significantly differ from typical age-matched healthy adolescents. The intervertebral efforts of one adolescent with IS and an age-matched adolescent during a complete gait cycle were computed and compared. **Results:** All intervertebral efforts are larger in the adolescent with IS compared to the healthy adolescent, except for the vertical torque. The average medio-lateral torque and force for the participant with IS are respectively 200% and 114% higher. **Conclusion:** This study revealed that the pathological efforts are not concentrated around critical points but distributed along the spine. Thus, higher average efforts along the spine in adolescent with IS may influence the spine deformity due to mechanical modulations according to the Hueter-Volkman Law. The potential of this model is promising for the therapeutic follow-up of adolescents with IS because it provides real-time efforts along the spine, as well as the corresponding information about the asymmetrical behavior of the spine during gait.

**Keywords:** Multibody Model, Inverse Dynamics, Idiopathic Scoliosis, Intervertebral Effort, Gait

## Introduction

Idiopathic scoliosis (IS) is a three-dimensional (3D) deformity of the spine and of the trunk affecting 2-3% of the adolescents<sup>1</sup>. The main challenge in treating adolescents with idiopathic scoliosis (IS) consists in constraining the progression. Recommended treatments for those adolescent is conservative; when curves are between 15 and 45° they are treated using physiotherapeutic scoliosis-specific exercises and bracing in curves above 20°<sup>2</sup>. However, surgery is often performed for adolescents with scoliosis higher than 45°<sup>3</sup>. The SOSORT guidelines<sup>2</sup> suggests early conservative treatment to avoid scoliosis progression and

possible invasive spinal surgery, as surgery can negatively affect physical performance, such as walking<sup>4</sup>.

Walking is one of the most common daily activities and with IS, the spinal deformation causes an asymmetry of the trunk which leads to gait abnormalities. A recent survey<sup>5</sup> reported that several studies have found differences between healthy adolescents and adolescents with IS for different parameters. For example, adolescents with IS present different gait patterns for temporal spatial parameters, cinematic parameters, kinetic parameters, mechanical work, and energy expenditure when compared to their healthy counterparts. Particularly, adolescents with IS show decreased antero-posterior and medio-lateral pelvis motion<sup>6,7</sup>, decreased medio-lateral and antero-posterior hip motion<sup>6,7</sup>, decreased antero-posterior hip motion<sup>8,9</sup>, and decreased antero-posterior knees motion<sup>7</sup>. The decreased range of motion in adolescent with IS might have an impact on the coordination between the thorax and the pelvis<sup>9</sup> during gait, meaning that the efforts of the lower limbs might be transferred differently to the spine. It is worth noting that, "efforts" refer to both forces and torques. The efforts applied to the spine have been found to significantly influence the

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progression of the Cobb angle in AIS. The “Hueter-Volkman Law” stipulates that vertebral growth is slowed by mechanical compression<sup>10</sup>. In fact, a mechanical compression reduces the number of new cells produced in the proliferation zone of the growth plate<sup>11</sup>. Asymmetric efforts applied on vertebrae growth plates on the concave side of the curve inhibit growth and enhance the spine deformation. Bracing, the typical non-surgical treatment for scoliosis, is based on this principle and attempts to unload the growth plates<sup>12</sup>.

Efforts have been shown different for adolescent with IS. Yazji et al.<sup>13</sup> demonstrated that the medio-lateral joint forces in the lower limbs are significantly different between adolescents with IS and age-matched controls during gait. Raison et al.<sup>14</sup> used a multi-body model to demonstrate that the efforts between the pelvis and the spine are different between an adolescent with IS and a healthy adolescent. Additionally, Yazji<sup>4</sup> has developed a tool to compute intervertebral efforts all along the spine during gait: these efforts are broken down into three components of forces and three components of torques, and that is for the 18 intervertebral discs between L5 and C7, leading to a total of  $(3+3) \times 18 = 108$  analysis data points. The intervertebral efforts in pre-operative adolescent with IS than those were more different of healthy adolescents in this condition. In the postoperative condition, intervertebral efforts in adolescents with IS were more similar to those calculated in healthy adolescents<sup>4</sup>.

Those results are interesting and the model could lead to a great tool for therapeutic follow-up. However, none of the above-mentioned studies consider the spine kinematic. Yazji<sup>4</sup> and Raison et al.<sup>14</sup> consider that the spine 3D reconstruction is based on bi-planar radiographs and stays fixed during gait. However, Schmid et al.<sup>15</sup> showed that spinal gait kinematics is critical for adolescents with IS, by demonstrating that the curvature angle range of motion during gait is more important in the thoracic and thoracolumbar direction for adolescents with IS than for healthy adolescents. Their results are aligned with the “Nottingham concept”, which stipulates that kinematic differences in the spine, pelvis, and lower extremities during gait might contribute to the progression of AIS<sup>16</sup>. Thus, to the knowledge of the authors, there is still no solution in the literature that assesses the efforts in real-time along the spine with the consideration of the spine kinematics. We believe that these efforts are a key element to the progression of the scoliosis, and that a better understanding of their behavior would support the rehabilitation process of adolescents with IS.

This research work is based on the hypothesis that a multibody model of the human body will make it possible to quantify the efforts along the spine and that these efforts may be different between an adolescent with IS and an age-matched healthy participant.

Therefore, the objective of this study is to develop a multibody model paired with the kinematic model of the spine. Intervertebral efforts will help to visualise the intervertebral efforts along the spine of adolescents with IS, and to highlight the efforts that significantly differ from those found on age-

matched healthy adolescents for therapeutic follow-up using. This dynamic model should be able to run in real-time for simultaneous display on a computer screen and should be usable on any type of posture and/or motion.

## Methods

### *Multibody model of the human body*

A real-time inverse dynamic model of the human body was developed to assess the intervertebral joint efforts (Q). The model is composed of 25 rigid bodies (the lower limbs (6), pelvis (1) and spine (18)) and is illustrated in Figure 1. The model is personalised to each participant with measurements taken directly on them by a physiotherapist, i.e. the participants' mass and height, the width of the knee and the ankle and the foot length. From those measurements, the center of rotation of each articulation is calculated based on the model of Davis III<sup>17</sup>. The body inertia parameters, i.e. the mass  $m$ , moments of inertia  $I$  and center of mass COM coordinates of each body, were computed from the inertia tables from de Leva<sup>18</sup> by using the total mass, the height, and the sex; finally, the inertial parameters of the vertebrae were computed following Kiefer et al.<sup>19</sup>.

Here is an order of magnitude of the reliability, accuracy, and sensitivity of the process, also considering the potential inaccurate placement of the sensors with respect to the anatomical landmarks, and the soft tissue artifacts during motion:

First, concerning the joint angles, which are the intermediary variables computed for the assessment of the joint efforts in Eq. 2.1, a systematic review from McGinley et al.<sup>20</sup> on the reliability of three-dimensional kinematic gait measurement and processing revealed that “most studies providing estimates of data error reported values of less than 5°”, showing “evidence that clinically acceptable errors are possible in gait analysis”. Especially with Vicon cameras and its Davis III's model<sup>17</sup>, on which our model was based to compute intervertebral efforts, the accuracy of the sensor coordinates is known to be around “3 mm” and “did not have a significant effect on the measured joint angle patterns”<sup>21</sup>. Concerning the sensitivity, let us note that the kinematic data are recognized as the most sensitive parameters of the process<sup>20</sup>, especially for both following reasons: on the one hand, errors on joint angles can be up to 10° if some sensors are misplaced<sup>22</sup>, which especially happens when the raters who place the sensors on subjects change<sup>23</sup> this is why we had only one rater for the subjects, and recommend to do so. On the other hand, joint angles (resp. joint efforts) are known to be influenced by skin motion artifacts, of which the errors are 20-25%<sup>24</sup>; this problem is usually solved by calculating the kinematics by global optimization<sup>25</sup>, exactly as we have performed and recommend it.

Secondly, concerning the joint efforts computed in the end, we think that the order of magnitude of accuracy and reliability is around 4%. Indeed, at the lower limbs, Monte Carlo simulations showed that the worst RMS torque errors

averaged 4% bodyweight  $\times$  height<sup>26</sup>; at the upper limbs, an analysis of the joint torque variability and repeatability showed a “good repeatability” of the peak torques and that “the variability coefficient of trial peak torques remained under 4%”<sup>27</sup>. Obviously, future studies would be necessary to validate the *in vivo* precision of these internal efforts in the human body. However, the reason why this precision has not yet been directly assessed and validated by sensors is that the currently known sensors are recognized as highly invasive, e.g. by placing between vertebrae some strain gauges<sup>28</sup> or pressure sensors<sup>29</sup>. These invasive sensors would also potentially affect the quality of the results, and are finally no more ethically recommended today, which justifies again the interest of using modeling to assess the internal efforts, such as was proposed in this study for the intervertebral efforts.

The lower limbs are modeled as 3 bodies (thigh, leg and foot) linked together by ball joints and attached to the pelvis. The spine is also modeled as a chain of 18 bodies linked with ball joints. The external forces applied to the spine were simplified, where only the gravity and the ground reaction forces transmitted by the lower limbs are taken into consideration. The muscle activity and the relative motion of internal organs were neglected in this model. The intervertebral efforts computed in this study result from different muscle activities, i.e. different muscular co-contractions. Indeed, Abedrabbo et al.<sup>30</sup> pointed out a broad impact of the muscular co-contraction on the intervertebral efforts. Let us remind that the intent of this study was to identify the interesting components of joint efforts among the 108 calculated efforts throughout the spine, which already represents a high number of variables analyzed in this discussion, without going into the details of muscle forces. But as Abedrabbo et al.<sup>30</sup> showed that these intervertebral efforts are indicators potentially revealing different levels of muscle co-contractions, this could be the topic of future studies. Note, however, that a musculoskeletal modeling will not be trivial at the spine level, including a major challenge: the solving of the muscle over-actuation problem around each joint, for which no process has seemed yet transferable in real-time, notably because of a necessary preliminary step of muscle force calibration, as presented by Amarantini et al.<sup>31</sup> at the lower limb, or Raison et al.<sup>32</sup> at the upper limb, weighing down the process both at the levels of the experimental protocol and the computation time.

The mass of each trunk slice at each vertebra level was taken into consideration, including the mass of the internal organs. But the relative movement of each internal organ was neglected. Indeed, Abedrabbo et al.<sup>33</sup> showed that organs movements, considering their anchorage points, do not increase the intervertebral efforts significantly. Our interpretation is that the dynamics of the internal organs is not so much significant during natural movements such as gait and we choose not to include them in our model.

The calculation of the intervertebral efforts is divided into two main stages repeated for each time step; the first stage concerns the kinematic identification of the whole human body including the spine and the second stage is about the

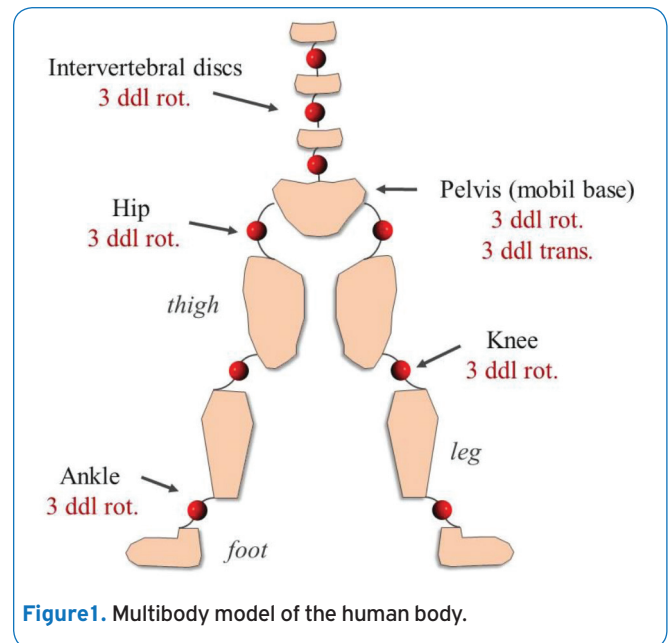


Figure 1. Multibody model of the human body.

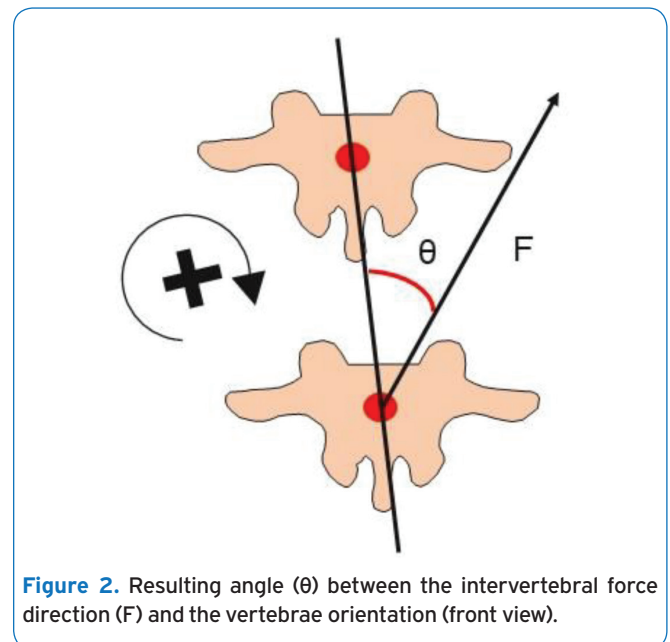


Figure 2. Resulting angle ( $\theta$ ) between the intervertebral force direction ( $F$ ) and the vertebrae orientation (front view).

execution of the inverse dynamic process. In the first stage, the recorded positions enable one to compute, in real-time, the model 78 relative coordinates (54 for the spine, 18 for the lower limbs and 6 for the pelvis) by inverse kinematics using the Levenberg-Marquardt optimization algorithm developed by ALGLIB® ([www.alglib.net](http://www.alglib.net), Sergey Bochkanov). The joint kinematics were obtained from the Cartesian coordinates of the anatomical landmarks, recorded by the cameras, using an inverse kinematic process combined with a global optimization<sup>25</sup>. While, the second stage provides the

**Table 1.** Mean values of intervertebral torques and forces in 3 different planes for A. a healthy participant, and B. a participant with IS.

	Torques			Forces			Force directions	
	antero-posterior [mN.m/kg]	medio-lateral [mN.m/kg]	vertical [mN.m/kg]	antero-posterior [mN/kg]	medio-lateral [mN/kg]	vertical [mN/kg]	antero-posterior [deg]	medio-lateral [deg]
<b>A.</b>								
L4-L3	90 (40)	0 (70)	-20 (30)	1720 (300)	400 (640)	2 790 (220)	20,6	16,6
L3-L2	60 (30)	20 (70)	0 (10)	630 (190)	20 (500)	3 020 (220)	7,4	3,4
L2-L1	40 (30)	30 (60)	0 (10)	680 (140)	90 (410)	2 800 (240)	13,2	3,4
L1-T12	30 (30)	30 (50)	10 (10)	20 (110)	-270 (360)	2 640 (220)	0,6	-6,9
T12-T11	30 (20)	20 (40)	0 (10)	390 (110)	-120 (300)	2 390 (210)	6,9	-0,6
T11-T10	20 (20)	20 (30)	0 (10)	90 (110)	-20 (250)	2 180 (180)	2,3	1,1
T10-T9	10 (20)	20 (30)	0 (10)	-120 (90)	90 (220)	1 980 (170)	-4,6	2,9
T9-T8	20 (20)	20 (20)	0 (10)	10 (90)	-170 (180)	1 790 (170)	0,6	-6,3
T8-T7	20 (10)	20 (20)	0 (10)	160 (100)	-160 (160)	1 630 (140)	3,4	-4,6
T7-T6	10 (10)	10 (10)	0 (0)	100 (90)	-130 (130)	1 490 (130)	1,7	-4,6
T6-T5	10 (10)	10 (10)	0 (0)	70 (70)	50 (110)	1 360 (120)	0,0	4,0
T5-T4	10 (10)	10 (10)	0 (0)	140 (70)	-110 (100)	1 230 (110)	2,9	-6,3
T4-T3	0 (10)	10 (10)	0 (0)	-30 (70)	-110 (90)	1 110 (100)	-2,9	-6,3
T3-T2	0 (0)	0 (0)	0 (0)	120 (60)	-20 (70)	990 (90)	0,6	0,0
T2-T1	0 (0)	0 (0)	0 (0)	50 (50)	-80 (60)	850 (80)	0,0	-7,4
T1-C7	0 (0)	0 (0)	0 (0)	40 (40)	-80 (50)	750 (70)	-2,9	-6,9
Total							49,3	-17,8
<b>B.</b>								
L4-L3	150 (70)	40 (50)	-40 (50)	3 040 (280)	220 (650)	2 290 (210)	35,0	6,9
L3-L2	90 (70)	60 (60)	-20 (50)	1280 (200)	240 (460)	3 310 (310)	9,2	5,7
L2-L1	60 (60)	60 (50)	10 (30)	120 (260)	-150 (340)	3 310 (250)	-1,1	-3,4
L1-T12	60 (50)	60 (50)	0 (10)	420 (260)	-120 (270)	3 030 (240)	7,4	-4,0
T12-T11	50 (50)	60 (40)	10 (10)	-200 (220)	-200 (240)	2 790 (220)	1,7	-6,3
T11-T10	50 (40)	50 (30)	10 (10)	240 (200)	-570 (220)	2 440 (210)	5,2	-17,2
T10-T9	40 (40)	40 (30)	0 (10)	140 (190)	110 (140)	2 290 (180)	2,9	10,3
T9-T8	40 (30)	40 (20)	0 (10)	340 (190)	-280 (140)	2 030 (170)	2,3	-12,0
T8-T7	30 (30)	30 (20)	10 (10)	200 (190)	-230 (130)	1 880 (150)	5,2	-6,3
T7-T6	30 (20)	30 (20)	0 (10)	140 (190)	-200 (120)	1 710 (140)	3,4	-7,4
T6-T5	30 (20)	20 (10)	0 (10)	150 (180)	-210 (120)	1 550 (130)	4,0	-9,2
T5-T4	20 (20)	20 (10)	0 (0)	180 (160)	-90 (100)	1 420 (110)	1,1	8,0
T4-T3	20 (10)	20 (10)	0 (0)	160 (150)	-150 (100)	1 270 (100)	6,9	-14,9
T3-T2	10 (10)	10 (10)	0 (0)	140 (140)	-100 (70)	1 130 (90)	6,9	1,7
T2-T1	10 (0)	10 (0)	0 (0)	170 (140)	-140 (70)	960 (*0)	-1,7	-17,2
T1-C7	0 (0)	0 (0)	0 (0)	120 (120)	-120 (60)	850 (70)	-1,7	-8,0
Total							86,5	-73,3

efforts based on a dynamical equations system obtained from a Newton-Euler formalism (Equation 1). ROBOTRAN software<sup>34</sup> was used to generate the symbolic equations.

$$\mathbf{Q} = f(\mathbf{q}, \dot{\mathbf{q}}, \ddot{\mathbf{q}}, \mathbf{F}_{ext}, \mathbf{g}) \quad (1)$$

The output variables are the efforts (forces and efforts) between each vertebra ( $\mathbf{Q}$ ). They are dependent of the joint angles ( $\mathbf{q}$ ), speeds ( $\dot{\mathbf{q}}$ ) and accelerations ( $\ddot{\mathbf{q}}$ ), the external forces ( $\mathbf{F}_{ext}$ ) and the gravity ( $\mathbf{g}$ ). Six efforts, i.e. 3 forces (N/kg) and 3 torques (Nm/kg), are calculated: the antero-

posterior, medio-lateral, and vertical directions.

To quantify the impact of the noise and the measurement errors on the computed efforts, we added a random noise to the Cartesian coordinates of the anatomical landmarks and we added the maximum error of the instruments on the measurements taken directly on the participants. The average of the differences between the 54 computed efforts along the spine and those computed with an additional noise on the Cartesian coordinates was 3,9% with a



**Table 2.** Average intervertebral torques and forces and its standard deviation (SD) along the spine in three different planes for a healthy participant and a participant with IS.

	Torques			Forces		
	Antero -posterior [mNm/kg]	Medio-lateral [mNm/kg]	Vertical [mNm/kg]	Antero-posterior [mNm/kg]	Medio-lateral [mNm/kg]	Vertical [mNm/kg]
<b>Healthy participant: mean (SD)</b>	20 (10)	10 (10)	0 (0)	160 (100)	-70 (100)	1 750 (700)
<b>Participant with IS: mean (SD)</b>	40 (20)	30 (20)	0 (10)	240 (300)	-150 (170)	2 000 (810)
<b>Relative difference: IS vs. healthy [%]</b>	100	200	0	50	114	14

standard deviation of 34%. The same average on the efforts computed with additional errors on the measurements and the standard effort was 1,0% with a standard deviation of 0.4%. Those results show that our model is not sensible the measurements errors, but it is still sensible to the noise on the Cartesian coordinates. Indeed, since we try to develop a real time algorithm we were limited in the options were limited to smoothen the signals. In that context, we think the sensitivity of our system is acceptable.

#### Experimental set-up

To test the validity of the model, a proof of concept was performed with one healthy adolescent and one adolescent with IS with a double scoliosis with a left thoracic curve of 29° and a right lumbar curve of 23°. They were both asked to perform a gait at a normal pace on a straight line for 5 meters. The acquisition system includes 54 optokinetic sensors based on the Plug-in Gait full body of Vicon on which 16 markers were added on the spine (vertebrae L5 to C7 were marked). The reliability of the Vicon model is known to be +/- 3mm and do not have a significant effect on the measured joint angles<sup>21</sup>. The tridimensional coordinates of the sensors were recorded by a 12-camera motion-capture system (Vicon, UK) to measure the movement. The ground reaction forces applied to each foot were captured from independent force platforms (AMTI, USA) placed in the middle of the walk. Data from one complete cycle were recorded for this study. The cycle started when the first foot touched a platform and ended when the second foot left the other platform.

All the data was synchronized at 100 Hz and was sent to our homemade C++ real-time routine to calculate the efforts in terms of forces and torques applied to each intervertebral joint.

## Results

Table 1 presents the average torques and forces in a gait cycle in three directions ( $x$ ,  $y$ ,  $z$ ) for each intervertebral joint for the healthy participant (A) and for the participant with IS (B). The efforts have been normalized according to the weight of each participant to limit its impact. The joint efforts

around both apex (L2 and T8), the vertebra that is located at the farthest point out laterally from the midline of the body, have been underlined. In addition, the force directions were calculated and compared to the line that would pass through the center of two adjacent vertebrae. The resulting angle ( $\theta$ ), illustrated in Figure 2, is presented in the last two columns of Table1. The total force deviation which is the summation of all the angles is also presented in the Table 1. The total deviation allows a quick understanding of the magnitude of this measure.

Table 2 presents the means of all the intervertebral efforts along the spine (all vertebrae) and the standard deviation (SD) along the spine. Referring to those results, the medio-lateral torque and force of the participant with IS are much higher compared to the healthy participant, respectively increased by 200% and 114%, respectively. The antero-posterior torque and force are 100% and 50% higher, respectively, for the participant with IS. The vertical torque is almost identical for both participants and the vertical force is a little higher (14%) for the adolescent with IS.

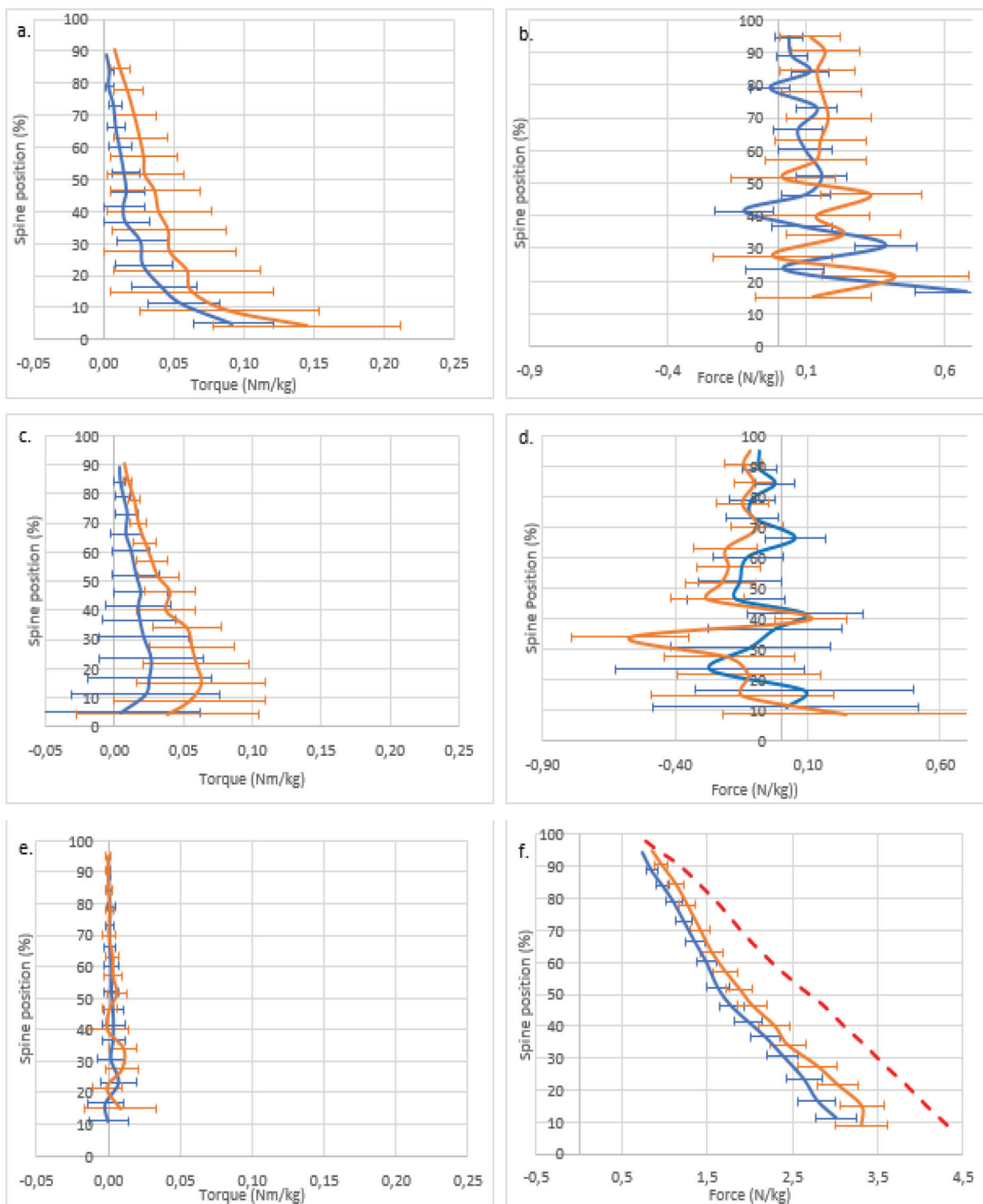
Figure 3 presents a graphical view of the effort distribution along the spine. The average value of each intervertebral effort during a gait cycle was calculated and associated with its position along the spine in percentages, the joint L4-L3 is considered the beginning of the spine (0%) and the joint T1-C7 is considered the end of the spine (100%).

## Discussion

### Multibody model of the human body

A multibody model of the human body was developed to assess the intervertebral joint efforts in real-time. Since there is no study that presented the intervertebral effort along the whole spine. The results were confirmed with expected values. First, the vertical force ( $\bar{F}_v$ ) around the lower discs and normalized to the participant mass, can be approximated by the equation 2, where the trunk and the head represent 49% ( $a$ ) of the participant total mass ( $m$ ) according to the adjustment table of de Leva<sup>18</sup>. It is represented on Figure 3f by the red line.

$$\bar{F}_v = \frac{m \times a \times g}{m} = 0.49 \times 9.81 = 4.8 \text{ N/kg} \quad (2)$$



**Figure 3.** Torques and forces distribution along the spine for a healthy participant (blue) and a participant with IS (orange) for a. antero-posterior torque, b. antero-posterior force, c. medio-lateral torque, d. medio-lateral force, e. vertical torque, f. vertical force. The red line on f. is the theoretical force distribution.

The vertical force on the top of the spine is expected to be around 0,69 N/kg since the head represents 7% of the total mass<sup>18</sup>. With our model, the vertical force between L3-L2 was 3,10 N/kg. The differences between the expected values and the measured ones are explainable by the fact that the force applied by the mass is distributed. For example, the antero-posterior force is more important in the lower part of the spine due to its natural lordosis in the lower back as we can see in the Table 1. The measured vertical force between T1-C7 is 0,89 N/kg which is a little higher of what was expected. However, one needs to consider that equation 2 is an approximation and it does not take into consideration the vertical movement of the head during gait. Additionally, the medio-lateral force acts as expected, as it oscillates around zero according to the natural balancing of the trunk during gait.

Finally, it would be interesting to compare our results with more complete models that would include muscle activities and masses of internal organs to validate our simplification hypothesis<sup>30</sup>. This could be done in a future work.

#### *Intervertebral efforts*

The Figure 3 shows the distribution of the efforts along the spine. The results presented show that the intervertebral joints around the apex are higher for the participant with IS, but there is not a peak as one would have expected if there was a concentration of efforts. The efforts seem to follow a logic linear gradation for both of the participants. The obtained results lead us to the conclusion that the spine should not be considered as one single vertebrae linked together but as a curve in tension. This means that a spinal deformation could not be equal to a concentration of efforts but to a redistribution of the efforts along the spine.

In Figure 3, we can also observe that their behaviors are similar but the efforts of the participant with IS are higher, i.e. less aligned with the spine. Table 2 shows that the summations of the efforts are more important for the participant with IS along four directions (antero-posterior torque and force, medio-lateral torque and force). By normalizing the efforts of the weight of the participant, it would be expected that they would be in the same order of magnitude. The noticed higher efforts for the participant with IS may be pathological and can be associated with asymmetrical behavior of the spine during gait.

Several studies<sup>10-11,35-36</sup> showed that mechanical efforts applied to vertebrae slow their growth based on the "Hueter-Volkman Law". Additionally, the direction of the forces presented in Table 1 being higher for the participant with IS (0,86 vs 1,51 rad for the antero-posterior force and -0,31 vs -1,28 rad for the medio-lateral force) means that the forces are less aligned with the spine curve. Moreover, the negative force direction indicates that medio-lateral forces are applied on the concave side of a left thoracic curve. This compression slows the growth of the vertebrae and enhances the deformity progression. The force directions could lead to a quantitative factor of the progression and could help to predict it.

#### *Clinical output*

The results from Tables 1 and 2 indicated that the numerical computation of internal efforts along the spine can be used as a qualitative support for a therapeutic tool for adolescents with IS. Future studies would be necessary to draw more quantitative conclusions, by assessing the precision and sensitivity of the global computation process, including potential errors in the geometrical measurements, potential discrepancies between literature-based mechanical properties and actual values, potential imprecision in the placement of the optical sensors, and soft tissue artifacts during motion. Therefore, this study supports the need for intervertebral quantification for the therapeutic follow-up of adolescents with IS.

## Conclusion

The objective of this study was to develop a multibody model that can assess intervertebral efforts for the development of a therapeutic follow-up tool for adolescents with IS. The obtained results in Tables 1 and 2 showed that the medio-lateral and antero-posterior torques and forces are different between an adolescent with IS and an age-matched healthy participant, and that they could be great indicators of asymmetrical spine behavior. The intervertebral force direction is a promising indicator for the progression of the deformity. Additionally, this study revealed that the pathological efforts are not concentrated around critical points, but are distributed along the spine.

Finally, it would appear too early to draw any categorical conclusion since this paper present results using only two participants. More research will be needed to confirm the generality of the results to translate them into therapeutic indicators. The next step is therefore to validate the developed tool with a larger cohort of adolescents with IS.

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