A FINITE ELEMENT MODEL STUDY ON THE ROLE OF TRUNK MUSCLES IN GENERATING INTRA-ABDOMINAL PRESSURE

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1. INTRODUCTION

The significant economic costs and human suffering due to work related low-back disorders have motivated many scientists from diverse disciplines to search for mechanisms of injury and the associated risk factors. Prevention strategies are predicated on the premise that risk factors can be identified, controlled or eliminated by appropriately designed interventions.

The nature of spinal loading (in terms of its magnitude, direction, repetition and duration) has been associated with the development and aggravation of work-related low back disorders (19-22,26-28,33,34,36,38). Hence, the spine has been the focus of many biomechanical evaluations on how external loads are carried and shared by the trunk muscles and osteoligamentous spine. Historically, the limitations in modeling of spinal musculature have led the investigators to search for auxiliary mechanisms assisting the extensor muscles in satisfying the equilibrium conditions (1,3,8,13,18,26-28,30). The detailed description of these theories and their biomechanical evaluations (testing the validity of claims regarding the unloading function of the intra-abdominal pressure and utilization of lumbodorsal fascia) can be found in a recent review by McGill and Norman (28). It is now recognized that the stability and equilibrium conditions of spine must be simultaneously insured by proper neuromuscular functions of various components of spine (1,4,7,8,12,15,17,33,33-36,40).

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While we have advanced in anatomical modeling of posterior spinal musculature due to reported pioneering investigations (2), we still lack a similar precision about the anterior trunk muscles despite some recent improvements (29,32,39). The transverse abdominis, internal and external oblique muscles are flat muscles. The line of action of oblique muscles also varies around the trunk. Hence, representing oblique muscles with a single equivalent muscle is likely inadequate from both mechanical and neural control perspectives (11). There is evidence that different portions of oblique muscles may be selectively activated (11). The role of transverse abdominis has been neglected in most equilibrium-based models since its horizontal orientation covering the anterior abdominal wall limits its direct moment generation capability about the lumbar spine. Its location beneath internal and external obliques has also made it inaccessible for the non-invasive surface electrodes favored in biomechanical studies.

Many have evoked the role of intra-abdominal pressure (IAP) in equilibrium considerations by unloading of spine and assisting the extensor muscles during lifting tasks (8). More recently, the concept of higher IAP contributing to the stability of spine by stiffening of spinal column has been proposed (1,2,40).

Clinical studies have shown that there are significantly higher delays in recruitment of transverse abdominis (TA) muscles during the self-generated postural perturbations that is caused during the movement of lower limbs (15,16,34) in low back pain patients. The activities of TA, however, precede the activities of other trunk muscles and the primary movers of the lower limbs in normal subjects. Based on assumed importance of neural control of TA, a special theory and treatment strategy for segmental stabilization of spine by trunk muscles has been developed with significant positive results in practice (34). A quantitative assessment of the role of TA in the spine mechanics is

never-the-less missing.

The aforementioned void motivated the development of the following model as a first step towards a comprehensive validated model of the human trunk. The muscles are modeled as continuous fiber-reinforced membranes that surround the intra-abdominal cavity. These membranes, unable to carry any moments, can resist tension primarily along their fiber direction; they develop in-plane stresses while generating IAP. The finite element model is used to allow the varying distribution, cross-sectional area and fiber direction of muscles around the trunk. Hence, this model improves upon the Daggfeldt and Thorstensson model (8) that considered only one fiber orientation and muscle type covering the whole trunk. While the model is still idealized due to its lack of passive spine and anatomical details, it is a first step towards a novel approach to modeling of abdominal muscles. It was thought that evolution of the model should take a systematic step to allow us to appreciate the contribution of each added element. The ultimate goal of this modeling effort is to allow realistic appraisal of the role of trunk muscles in maintaining the equilibrium and stability of spine under normal daily and occupational activities as well as clinical conditions.

2. METHOD

2.1 Anatomical Description of Trunk Muscles

and Intra-abdominal Cavity

The abdominal wall was assumed to be a cylindrical membrane (8) with a radius of 6.78 cm based on the average peritoneal cavity of 144.4 cm² at the L4/L5 (27). The height of cylinder has been considered as the breath of transverse abdominis that is about 11cm based on a cadaver study (27). The six muscle groups surrounding the abdominal wall are the erector spinae (ES), latissimus dorsi (LAT), rectus abdominis (RA), internal obliquus (IOB), external obliquus (EOB), and transverse abdominis (TA) (Fig. 1). The appropriate thickness was computed for each muscle based on their anatomical cross sectional areas (Table 1) and their circumferential distribution (Fig. 1). Fibers orientation for RA and ES has been taken in the longitudinal direction and in the transverse direction for TA whereas that for IOB, EOB and LAT has been selected from the study of Nussbaum et. al. (32) as shown in Fig. (2). Hence, each muscle has its own specific thickness, circumferential distribution and fibers orientation.

2.2 The Finite Element Model

A finite element (FE) model has been developed for assessing the effect of activation in each muscle individually or in combination on the IAP (Figures 3 and 4). The model was developed and analyzed using ANSYS 5.4 (ANSYS Inc., Southpointe, 275 Technology Drive, Canonsburg, PA, USA). The intra-abdominal cavity was assumed to contain a nearly incompressible fluid (with the bulk modulus of 250 MPa) surrounded

Fig. 1- Arrangement of muscles around the intra-abdominal cavity. The thickness and distribution of each muscle is based on the literature cited in Table 1.

Table 1. Anatomical cross sectional area (ACSA) and computed thickness of trunk muscles

Muscle	ACSA (cm^2)	Computed Thickness (mm)	Reference #
Rectus Abdominis (RA)	20	23.94	5
Latissimus Dorsi (LAT)	2x4	2.474	6
Internal Obliquus (IOB)	2x9	6.496	6
External Obliquus (EOB)	2x10	6.641	6
Erector Spinae (ES)	45.1	33.882	27
Transverse Abdominis (TA)	5	4.545	29

Fig. 2 Angle of fibers of IOB, EOB and LAT taken from Nussbaum et al. (1995). The ES and RA are aligned vertically while TR is aligned horizontally.

by three layers of membrane (Fig. 3). Each layer represented a muscle with transverse isotropic material properties with greater stiffness (Young's modulus) in fibers direction (37). Thickness of each layer and material principal directions (x-axis being in fibers direction) have been selected as given in Table 1. For example TA is a membrane layer with thickness of 4.545 mm and greatest stiffness in transverse direction and RA is modeled as 3 layers with total thickness of 23.94 mm with primary stiffness in longitudinal direction. The three flat muscles (IOB, EOB, TA) have been represented with three overlapping layers with specific fibers direction and thickness. Also ES is modeled as two layers with total thickness of 33.882 mm and LAT as one layer covering the ES (Fig. 4).

The top plate simulates the active and/or passive roles of diaphragm through its vertical displacement. When the plate moves downward, the cavity-filled fluid offers resistance and IAP increases generating

forces in surrounding muscles. The membrane element is a spatial 4-node plane stress element having in-plane stiffness but no bending (out-of-plane) stiffness. The element has three degrees of freedom at each node (translations in the x, y and z directions). The cavity-filled fluid element is modeled by eight-node elements (3D contained fluid elements with no net flow) having three degrees-of-freedom at each node (similar to membrane elements). Brick elements have been used to model the solid structure of the top plate. Fluid nodes at the boundary coincide with the membrane nodes and are coupled to them only in the radial direction normal to the interface. The nodes at the bottom surface of the top plate are coupled to fluid nodes in the longitudinal direction. Hence, when the plate moves downward and the IAP increases, an upward force is generated acting on the diaphragm. All three layers of membrane elements are connected to each other at nodes by radial-oriented links allowing uncon-

Fig. 3a

Fig. 3b

Fig. 3 The FE model of intra-abdominal cavity. The model consists of nearly incompressible fluid contained by three layers of membrane simulating the muscles of the abdominal wall. The top plate displaces vertically to simulate the role of diaphragm.

strained relative motions along circumferential and axial directions. The top and bottom membrane nodes are fully constrained. The top plate is displaced downward at a prescribed magnitude causing membrane force/stress in the muscles, hydrostatic pressure (IAP), and constraint (reactive) force on the top plate.

2.3 Material Properties

Fiber layers of muscles are considered as membranes with direction-dependent material properties. The transverse isotropy is assumed in the plane normal to the direction of fibers. Each membrane has five independent moduli: the Young's modulus of membrane in the fibers direction (E_x), in the transverse directions ($E_y = E_z$), the shear modulus ($G_{xy} = G_{xz}$), and two

Fig. 4 Three membrane layers representing the trunk muscles.

Poisson's ratios ($\nu_{xy} = \nu_{xz}$ and ν_{yz}). For the inactivated muscle, the following material properties is considered (12):

$$E_x = 0.4 \text{ MPa}$$

$$E_y = E_z = E_x / 3$$

$$G_{xy} = G_{xz} = E_y / 2$$

$$\nu_{xy} = \nu_{xz} = 0.45$$

$$\nu_{yz} = 0.75$$

2.4 Simulations: Quiet Standing and Valsalva Maneuver

The IAP in the quiet standing position has been reported to be 0.98 KPa (31). When all muscles are inactivated, the top plate is moved downward until the IAP reaches 0.98 KPa predicted at 3.15 mm displacement. This position of the top plate was kept constant for all our simulations in this investigation. Maximum IAP has been considered 3.95 KPa for partial valsalva maneuver (12). Considering the muscles as variable stiffness elements, the Young's modulus along all muscle fibers was subsequently increased until the IAP reached 3.95 KPa. This IAP was generated at $E_x = 1.612 \text{ MPa}$ while remaining moduli were scaled accordingly with proportions stated in the previous section. This computed value is referred to as E_v .

Fig. 5 Predicted variation of IAP (Pa) and the upward (constraint) force (N) that acts on the top plate versus β ; the relative activation of individual muscles

2.5 The Effect of Variation in Individual Muscle Activation on the IAP

For determining the effect of each muscle on the IAP, a parameter β was defined as:

$$\beta = \frac{E}{E_v} \quad \text{for } \beta_{Min} < \beta < 1 \quad (1)$$

where E is the Young's modulus in fibers direction of each muscle and E_v is the upper-limit value that was obtained simulating the valsalva maneuver (1.612 MPa); $\beta_{Min} = 0.248$ calculated as the ratio of modulus for the inactivated muscle (lower-limit value of 0.4 MPa) and the E_v . Hence, β is changed from β_{Min} to 1 for different muscles, individually or in combination, and the analysis is performed for each value of β . The value of β indicates the activity level of muscles relative to that during valsalva maneuver (i.e., the upper-limit). It must be noted that none of the muscles were found to reach their full activation capacity during valsalva maneuver (31). Additional cases were also performed to study the effect of changes in muscle fibers direction on the IAP for the three flat muscles at $\beta = 1$. The muscle membranes were all verified to remain in tension in all directions.

3. RESULTS

The IAP increases with increase in activation level, β , for each muscle (Fig. 5). The muscles in the

Fig. 6 The unloading effect of IAP due to various activation levels of TA, IOB and EOB.

descending order of their efficiency to generate IAP at $\beta = 1$ are predicted to be TA, IOB, EOB, RA, ES, and LAT. The upward constraint force at the top plate shows similar trends for different muscle activation levels exhibiting a nonlinear response for muscles with longitudinal fibers direction (RA and ES) (Fig. 5). The direct unloading force due to higher IAP can be quantified as the difference between the foregoing constraint force and the longitudinal component of the muscle fibers force. It is shown that only TA can create an unloading force due to its horizontally directed muscle fibers (Fig. 6). The rest of muscles generates longitudinal forces greater than the unloading force they create by their contribution through the IAP. The maximum fibers stress (at $\beta = 1$) remains well below the maximum permissible stresses in all muscles (Fig. 7).

The complexity of the effects of muscle functional anatomy and architecture on the IAP generation capability is shown in Fig. 8. While the EOB and IOB, if with fibers aligned horizontally, would generate more IAP than TA (due to their larger cross sectional area), they are less efficient in their more realistic fiber angles.

The effect of different synergistic recruitment of muscles in generating IAP is depicted in Fig. 9. It is clear that a desired level of IAP can be achieved by different recruitment patterns. However, the compressive penalty of these different recruitments will significantly vary. It is observed that the TA (Fig. 6) is the only single muscle that contributes to a net upward

Fig. 7 Tensile stress in fiber direction for different levels of activation in individual muscles.

longitudinal force while the other muscles individually or with synergistic recruitment would likely add to net compressive load on the spine. The other observation that is worth considering with caution is the inability of some combination of muscles (e.g. LAT and ES) to raise sufficient IAP to match the IAP generated by the TA alone. The combined activation of IOB and EOB at valsalva activation level only slightly surpassed the IAP produced by TA alone (Figs. 5 and 9). The clinical implications of these observations must await till a more realistic representation of the passive spine is included and the biomechanical fidelity of the model is enhanced.

4. DISCUSSION

The idea that pressure within the trunk might assist unloading the spine was suggested in the 1920s (18). The underlying theory is that when, for example, a weight is lifted and a flexion moment develops about the torso, this moment is primarily counterbalanced by the posterior back muscles. The pressure in the trunk cavities assists in this respect, producing an extension moment. Two factors responsible for over estimation of IAP role has been the large cross sectional area of diaphragm and the large moment arm of the net upward force acting on the diaphragm (28). The FE approach in this study obviates the need to assign a priori a moment arm to the net upward force which are computed as the displacement at the top plate is prescribed.

Due to the fact that higher IAP is present when the spine is subject to different external moments, the role of IAP must be more complex than the reduction

Fig. 8 Variation in IAP (Pa) as a function of fiber angle from transverse (90°) to longitudinal (0°) orientations.

of the extensor moment on the spine (14). Indeed, TA activity is observed with different levels and directions of external moments (34). An increase in the IAP is commonly observed when a large load is placed on the spine during daily activities like lifting, jumping and running (3,9,10,23,26). The function of this increase in pressure is still not well understood. McGill and Norman (28) stated that the possible unloading of the spine by an increased IAP would be counteracted by an equally large or even larger loading by the activated abdominal musculature. On the other hand, some other investigators (3,8,40) claim that IAP can reduce the compression load on the spine. Currently, IAP is generally thought to provide stability to the lumbar spine (4,7,25). The co-contraction together with elevated IAP can effectively stiffen the spine to achieve required stability level (4, 26).

The purpose of the present study was to establish a FE model to illustrate the effect of trunk muscles, individually or in combination, on the IAP. The force acted on the diaphragm and resulting stresses and forces in the muscles have also been calculated. In previous studies (22,32,38), each muscle was considered as a single force vector. Obviating this deficiency, we considered each muscle as a membrane layer with greater stiffness in the fibers direction. To assess which muscle can unload the lumbar spine, we followed the method of Daggfeldt and Thorstensson (8). They suggested that for unloading the lumbar spine the

Fig. 9 The effect of different combinations of muscle recruitment patterns at $\beta=1$ on the generated IAP (KPa).

beneficial effects of increasing the IAP (computed constraint force, F_C) need to be greater than the negative effects due to the increased longitudinal tension in the intra-abdominal wall. The difference between F_C and F_L (increase in longitudinal tension in wall due to activity of the same muscle) is shown in Fig. 6. Longitudinal tension has been calculated from reaction forces on the fixed nodes at the top end of membrane layers. Results show that EOB cannot unload the lumbar spine. Daggfeldt and Thorstensson (8) in their simple model of intra abdominal cavity stated that muscles with fibers direction greater than approximately 55° with the vertical will contribute to unloading while tension in muscles with fibers direction smaller than 55° will load the lumbar spine. In our model IOB has 42.4° fibers direction and is more efficient in generating IAP than EOB with smaller angle. It may unload the spine at very low activation levels and achieves almost very little compressive penalty at higher activation levels. However, TA remains the *only* muscle to unload the lumbar spine at all activation levels considered.

The relatively simple model presented here has a number of limitations which will be delineated after a brief review of functional anatomy of intra-abdominal cavity based on Tortora (41). This section is included to highlight what significant structures are lacking in the current model and to point towards future developments. The posterior abdominal wall consists of the lumbar vertebrae, parts of ilia, psoas major, iliocostalis and quadratus lumborum muscles. The anterolateral abdominal wall is composed of fascia and four pairs of muscles: EOB, IOB, TA and RA. The first three muscles are flat muscles whereas the last one is a straplike vertical muscle. The fibers of EOB run medially and inferiorly and those of IOB run laterally and inferiorly (almost at a right angle with EOB) while most of the fibers of the TA muscle are horizontally oriented

around the abdomen. The RA extends from the pubic crest and pubic symphysis to the cartilage of fifth to seventh ribs and xiphoid process. The aponeuroses of the EOB, IOB and TA muscles form the rectus sheath that encloses the RA muscles to join a tough, fibrous band called linea alba (white line) that extends the xiphoid process of sternum to the pubic symphysis. The moment arms of these muscles have been augmented (29) since they are greatest at the point of junction to form the linea alba.

These anterolateral abdominal muscles have the following functions: to contain and protect abdominal viscera; to sagittally flex, laterally bend and axially rotate the spine; to compress the abdomen during forced expiration, and providing the force for defecation, urination and childbirth. The anterolateral abdominal wall can contract and distend much more than the relatively bulkier and more stable posterior abdominal wall. Ventilation (inspiration and expiration) occurs by changing the size of the thoracic cavity. The most important muscle in ventilation is the diaphragm which with its dome-shaped musculotendinous shape separates the thoracic and abdominal cavities. The two parts of diaphragm are the central and peripheral portions. During the inhalation (inspiration) the central (tendinous) dome region moves down while the peripheral (muscular portion) is attached to the lumbar vertebrae, sternum, and costal cartilages. The additional functions of downward movement of central dome of diaphragm include pressurization of intra-abdominal cavity in conjunction with the anterolateral abdominal muscle, and assist in returning to the heart the venous blood passing through the abdomen.

It is clear that the present downward direction of the top plate to simulate the action of diaphragm is a gross simplification. The insertion of diaphragm muscle to the passive spine and the rib cage must be properly considered to create more realistic boundary conditions for future developments. The presence of challenges due to breathing or holding of the breath should also be considered for full appraisal of their biomechanical consequences as studied experimentally (23-25).

In short, the model developed has the following limitations and its results must be used with caution:

- Lack of passive osteoligamentous spine to share the external load transmitted to spine (e.g. during lifting); consequently lack of attachment points by which the internal loads are transmitted to the spine. Local muscles specially the multifidus must be included for evaluation of the local and global stability of spine.
- Lack of physiological boundary conditions for the floor and ceiling of abdominal cavity and attachments of diaphragm.

- Idealization of geometry as a cylinder.
- Lack of consideration of different fiber directions in each of muscle groups. It is known that the fibers of these flat muscles (IOB and EOB) may vary across the abdominal wall.
- Imprecise material properties which could influence the predictions.

Future models would need to include the passive spine and its proper interface with the abdominal and trunk muscles so that the external loads would be balanced by the proper activation of these muscles and full assessment of consequence of alternative recruitment patterns could be considered. The method of activation of the muscles may take more direct approaches such as optimization, EMG-driven modeling or kinematics based FE modeling (5,6,35,36,38). The absolute value of predictions could change when the model is refined but it is likely that the relative efficiency of TA shall remain valid. The serial CT scans or MRI should provide more physiological anatomical geometry to be included in future developments. Authors present this paper only as a first step towards providing justifications and feasibility to undertake more advanced studies. Subtle changes in the motor control of low back pain patients and their effects on IAP and alteration in temporal and amplitude patterns of muscles can only be assessed by complex models that could accurately represent the interactions of the neural control, passive and active spinal sub-systems.

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