CFD MODELING OF PRESSURE TRANSIENTS IN THE SPINAL CANAL DURING WHIPLASH MOTION – A PILOT STUDY

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ABSTRACT

The objective of this study was to make use of recent developments in Computational Fluid Dynamics (CFD) modeling to investigate the pressure phenomena and the hypothesized concomitant vein blood flow that take place in the spinal canal during extension and flexion motion of the cervical spine.

Keywords: Whiplash, Neck, Spine, Pressure, Biomechanics

Sagittal neck bending causes a length change of the spinal canal with a concomitant inner volume change. This would cause transient pressure changes in the cervical spinal canal during whiplash trauma (rapid injurious bending of the neck due to inertial loading as the torso is accelerated). The pressure may load the nerve root region due to pressure gradients along each intervertebral canal. Svensson et al. (2000) hypothesized that, during sagittal bending, blood volumes in the venous plexus must move along the spinal canal and through the vein bridges to compensate for the volume change. The pressure transient was verified together with signs of cervical spinal ganglion nerve cell body membrane dysfunction in animal experiments (Svensson et al., 2000). Eichberger et al., (2000) recorded pressure transients in post mortem human subjects (PMHS) and Schmitt et al. (2002) were able to recreate the pressure transient in a CFD model. Taylor et al. (1998) reported interstitial hemorrhage in cervical dorsal root ganglia in an autopsy study of victims who had sustained severe inertial neck loading.

The ganglion cells were considered to be of particular interest since they are associated with afferent nerve paths that are related to several typical whiplash symptoms (e.g. neck and shoulder pain, cervicogenic headache) and they are also a key link in central nervous system pain sensitization.

Based on the pressure transient findings a neck injury criterion NIC was developed by Boström et al. (2000). Boström’s model focused on the early s-shape motion of the cervical spine during a whiplash exposure, where the acceleration and velocity of the vertical blood flow along the spinal canal was hypothesized to make the major contribution to the recorded pressure transients. The NIC value is a combination of the relative horizontal acceleration and the relative horizontal velocity, between T1 and the occipital joint. It is based on the geometrical relation between the length change of the spinal canal during the early s-shape motion and the horizontal motion between T1 and the head. The NIC was later evaluated using crash pulses from real-world accidents (Eriksson, 2006 and Linder et al., 2004).

The range of symptoms that are associated with whiplash trauma to the neck are well documented. The exact location of an injury is, however, still unknown. Different injury sites and injury mechanism have been presented, and the cervical spinal ganglia are today one of the more interesting potential injury sites (Siegmund et al., 2009).

The aim of the present study was to implement a CFD model of the venous plexus with reasonable boundary conditions. The model would enable a more detailed understanding of the flows and pressures in the spinal canal during whiplash trauma.

METHOD

A simplified 3-D geometry of the human cervical internal venous plexus was implemented in the CFD software Fluent (Liu and Yang, 2008). It consisted of one vertical pipe, representing the internal venous plexus, and seven narrower lateral pipes representing the intervertebral vein bridges (Fig. 1). This model could replicate the bending and the length change of each segment of the spinal canal.

The model geometrical dimensions were derived from the literature. There was a lack of reliable properties of the fluid and the vessel walls. These properties were thus varied to study their influence. The
radial flexibility of the human spinal canal could not be modeled. This was instead simulated by fluid compressibility (Goeury, 2008) using a bulk modulus of $6.13 \times 10^5$ Pa. The CFD model was meshed into 53573 cells. The blood flow was simplified to laminar Newtonian flow, fluid density 1050 kg/m$^3$ and viscosity 0.0035 kg/(m*s). The top boundary condition (the foramen magnum) was varied to assess its influence. There are no valves in the venous plexus vessels. Thus the fluid was allowed to move freely in both directions. Blood flow upward along the vertical vessel and outward through the side pipes was defined as positive. Rear impact (delta-v 8 km/h) volunteer neck motion data of Ono et al. (2000) were adapted using a spline function in MATLAB (Fig. 2).

RESULTS

Fig. 3 shows eight calculated stages of whiplash motion. It indicates that the lower neck flexed slightly in the first 50ms, and then formed an s-shape during 50-100 ms with a maximum at around 70ms. Thereafter the neck went into extension. Fig. 4 shows the spinal canal length change at each level.

With a selected combination of parameter values, the output pressure patterns became qualitatively similar to earlier experimental data. Fluid properties such as bulk modulus (simulating radial flexibility of the spinal canal) and pipe end conditions turned out to have significant influence on the results. A bulk modulus of $6.13 \times 10^5$ Pa was chosen based on comparisons to properties of arterial walls (Goeury, 2008). Figure 5 shows the pressure response at the different spinal levels in the model and compared to the experimental findings of Svensson et al. (2000) in Figure 6. The initial dip in Svensson’s results (25 ms) was probably due to the fact that the experiment was started in a slightly flexed neck posture to maximize
the whiplash trauma. Svensson’s tests were far more severe than the volunteer tests of Ono (2000), so the pressure magnitudes should be far lower in the CFD model. The early transients in Figure 5 are expected to result from non-physiological boundary conditions at 0 ms including initial velocity (Figs. 9 and 10). The flow accelerations and velocities are displayed in Figures 7-10.

The boundary conditions at the top and bottom end were set here to loss coefficients of 0.1 and the intervertebral pipes had loss coefficients of 10. These combinations produced a pressure pattern that qualitatively resembled the experimental results. The highest pressure magnitudes occurred in the middle of the cervical spinal canal in accordance with the experiments. Variations in top boundary condition as well as in side pipe length influenced the pressure pattern greatly, both in the timing and in the spinal level of the deepest pressure drop.
DISCUSSION

A CFD model of the internal cervical venous plexus and the intervertebral vein bridges was implemented in the software Fluent. Motion data from Ono et al. (2000) were used to define the vertebral motions and concomitant volume changes inside the spinal canal. These volume changes drove the flow in the model and transient pressure patterns from earlier whiplash experiments on animals and PMHS could be simulated with a suitable selection of the model flow properties.

The acceleration patterns of Figures 7 and 8 show a direct relationship between the vertical and lateral flow components. The sharp flow acceleration peak at about 60 ms with the highest magnitudes at C1 to C2 levels, and the simultaneous dip at C7, indicate the importance of the inertial effects in the negative pressure transient. The fluid acceleration at the top and bottom ends of the spinal canal drives the characteristic, and potentially injurious, pressure dip in the middle of the neck (C3 and C4). This inertial effect was the basis for the NIC criterion.

This pilot study shows that it is possible to qualitatively recreate the pressure transients in the spinal canal in a CFD model and create a picture of how the local transient flows may look. A further refinement of the model is possible if better data on vessel dimensions becomes available together with improved data on blood and blood vessel flow properties. New experimental data that combine pressure measurements and detailed vertebral motion data would be very helpful.

If the ganglion injuries turn out in the future to have a proven relation to whiplash symptoms and the causal relation between pressure loading and injury is fully established, this type of a CFD model could become very useful. The model could tentatively function as a replacement for a traditional injury criterion. Neck motion data from crash tests with bio-fidelic crash dummies or results from human body models could be fed into the model to predict the risk of injurious pressure loading to the ganglia.

REFERENCES


