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A fluid-structure interaction model of the internal carotid and ophthalmic arteries for the noninvasive intracranial pressure measurement method

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ABSTRACT

Accurate and clinically safe measurements of intracranial pressure (*ICP*) are crucial for secondary brain damage prevention. There are two methods of *ICP* measurement: invasive and noninvasive. Invasive methods are clinically unsafe; therefore, safer noninvasive methods are being developed. One of the noninvasive *ICP* measurement methods implements the balance principle, which assumes that if the velocity of blood flow in both ophthalmic artery segments – the intracranial (IOA) and extracranial (EOA) – is equal, then the acting *ICP* on the IOA and the external pressure (*Pe*) on the EOA are also equal.

To investigate the assumption of the balance principle, a generalized computational model incorporating a fluid-structure interaction (FSI) module was created and used to simulate noninvasive *ICP* measurement by accounting for the time-dependent behavior of the elastic internal carotid (ICA) and ophthalmic (OA) arteries and their interaction with pulsatile blood flow.

It was found that the extra balance pressure term, which incorporates the hydrodynamic pressure drop between measurement points, must be added into the balance equation, and the corrections on a difference between the velocity of blood flow in the IOA and EOA must be made, due to a difference in the blood flow rate.

1. Introduction

Today, severe traumatic brain injury (TBI) is one of the leading causes of disability and death worldwide, especially among children and young adults. Each year, around 7 million cases of TBI are recorded, and it is calculated that by the year 2020, this threat will become number one among fatal injuries [1]. Intracranial and central nervous system tumors are the 17th most common cancer type worldwide, with more than 256,000 new cases diagnosed in 2012 [2]. Infections after cranial surgery are also very serious threats that require immediate recognition and treatment [3]. Microgravity conditions could affect the human body's fluidic system, which in turn affects intracranial pressure (*ICP*), leading to vision deterioration or even vision loss [4]. In these cases, adequate patient monitoring is required to prevent secondary brain damage and to select the available treatment option in a timely manner. After experiencing TBI, or in a case of an intracranial tumor, a patient's brain begins to swell; this swelling could be monitored indirectly by measuring *ICP*.

ICP is the pressure inside the human skull and thus in the brain tissue and cerebrospinal fluid. The increase in *ICP* could be acute or chronic [5]. For a healthy patient in the supine position, normal *ICP* values are in the range between 7 and 15 mmHg [6], and for pathological patients, *ICP* can exceed 25 mmHg [7]. Pathways leading to an increase in *ICP* are due to intracranial tumors, blood vessel anomalies, infections, TBI, etc.

A human skull, once the sutures and fontanelles have closed, becomes a structure that permits no further expansion; consequently, the dependence of internal volume on *ICP* is negligible [8]. *ICP* is a result of interactions among internal constituents such as arterial blood, venous blood, cerebrospinal fluid, and brain tissue. Brain tissue is sensitive to blood flow dynamics, and several mechanisms (metabolic, myogenic, and neurogenic) are involved in maintaining the appropriate cerebral blood pressure. Within certain limits, a human body utilizes autoregulatory mechanisms of *ICP*, which is also known as the Monro-Kellie hypothesis [9]. When the autoregulatory mechanism fails, secondary brain damage may occur.

The gold standard for *ICP* measurement is the use of intraventricular catheters that are connected to an external pressure transducer; nevertheless, this invasive method greatly increases the risk of complications [10]. Therefore, an accurate and certified noninvasive method of *ICP* measurement is needed. Although there are several proposed noninvasive *ICP* measurement methods, such as numerical modeling, medical imaging, the implementation of the impedance mismatch principle, etc. [5,11,12], our study only focuses on the

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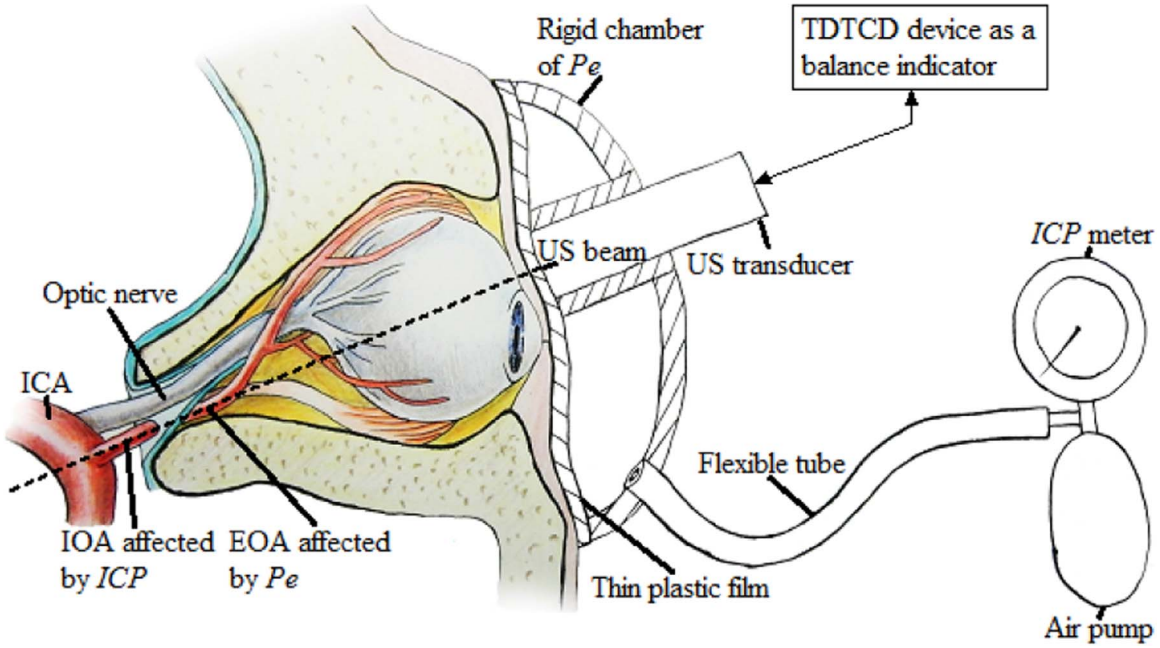

Fig. 1. A schematic of the noninvasive *ICP* measurement. A special mask that consists of a rigid chamber with a deformable thin plastic film and an ultrasonic (US) transducer is placed on the patient's head. A connection is formed between the chamber and air pump through a flexible tube. The air pump is used to increase the air pressure inside the chamber of the mask; consequently, the thin plastic film surrounding the eyeball begins to deform, generating *Pe,* which affects the EOA segment. Simultaneously, while the *Pe* is being applied on the EOA segment, the US transducer measures the velocity of blood flow in the IOA and EOA segments and sends the data to a two depth transcranial Doppler (TDTCD) device.

noninvasive *ICP* measurement method, which implements the balance principle that works without any calibration needed and uses the ophthalmic artery (OA) as the main sensor [13,14] to estimate the value of *ICP*.

OA supplies oxygenated blood to the eye. In most cases, it is the first intracranial bifurcation of the internal carotid artery (ICA), which in turn arises from the common carotid artery, which bifurcates from the aorta. OA starts inside the cranium and traverses the optic nerve canal to the eye socket located outside the skull [15] (Fig. 1).

The noninvasive *ICP* measurement method is based on simultaneous measurement of blood flow velocity in the OA at two points using the two depth transcranial Doppler ultrasound (TDTCD) technique: in the intracranial space of the OA (IOA) and in the eye socket, where the extracranial segment of OA (EOA) lies, by introducing an additional external pressure *Pe* in the cushion, tightly enclosing the area around the eye by placing a special mask on the patient's head. The mask surrounds the eyeball so that *Pe* can be applied on the eye, as also on the extracranial segment of the OA, while the IOA segment is affected by *ICP* (Fig. 1). At any given moment, the blood flow rate in different arterial segments is equal, but velocity differs, as it depends on the blood flow resistance and OA lumen diameter, which is influenced by *ICP*, *Pe,* mechanical properties of the OA, and the surrounding environment (tissue, fluids). In most cases, the pressure in the eye socket is zero (i.e., equal to atmospheric pressure). Because OA is affected by *ICP* inside the cranium, the diameter of OA inside the cranium is smaller than that on the outside of the cranium, and the respective blood flow velocity is larger. As the external pressure *Pe* in the mask enclosing the face increases, the diameter of the segment of the artery located in the eye socket decreases and the velocity of blood flow increases. The velocity of blood flow in both segments becomes close when the added *Pe* becomes close to the *ICP*. The condition of *ICP* and *Pe* balance is described by the following equation:

$$ICP = Pe + \Delta Pe_k \tag{1}$$

where ΔPe_k is the extra balance pressure term incorporating the influence of the hydrodynamic pressure drop (between the measurement points) of blood flow Δp_{blood} and other sources, such as any constrictions in movement due to bifurcation or prescribed displacements, material model, calculations errors etc., which affect the balance principle.

It is assumed that the OA's mechanical properties do not vary longitudinally because of its short length (~3 mm), but circumferential mechanical properties vary because of the OA's structure, which can be divided roughly into three parts: the tunica intima, tunica media, and tunica adventitia. Tissue or fluid enveloping OA influences its ability to change the diameter. During *in vivo ICP* measurements, residual blood velocity depends on the artery's mechanical behavior as well as all the other enclosing (tissue, fluid) properties and resistance to flow. The measured blood flow velocity values are used to calculate the comparative nondimensional indices, for example, PI (pulsatility index), RI (resistivity index), etc.

A working device for noninvasive *ICP* measurement was created and is already patented [13,16,17]. The device was practically tested and validated through *in vivo* TDTCD measurements and the results compared with those obtained using the invasive methods [14,18,19], including the "gold standard" ventricular *ICP* method [20]. In order to use the device in clinical practice, the accuracy of noninvasive *ICP* measurement must be improved, because according to the ANSI/AAMI standards, the error of the *ICP* measuring device should not exceed ±2 mmHg in the range of 0–20 mmHg, and in the range of 20–100 mmHg, the maximum error should not exceed ±10% [21]. That requires a fundamental understanding of the blood flow in the OA and its interaction with the surrounding tissues when the EOA wall is exposed to varying external pressure *Pe*. Experiments are too expensive or even impossible, while numerical simulation of blood flow in OA in situations close to those encountered during the measurements will allow us to estimate the factors that influence the measurement accuracy.

The main objective of the present work was to create and validate a numerical model suitable for investigations of the noninvasive *ICP* measurement method.

A three-dimensional FSI numerical model was created that incorporated the ICA segment, just after the bifurcation from the common carotid artery, then following the bone segment and the inner cranial